Electromyographic Analysis of Hip and Knee Exercises: a Continuum from Early Rehabilitation to Enhancing Performance

By

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Abstract

Introduction: Muscles around the hip and knee regions work in unison within the kinetic chain to produce functional movements. After a musculoskeletal injury, a progressive programme of rehabilitation exercises should be completed in order to return the athlete to full function.

Aims: The primary aim was to identify a progressive continuum of lower limb exercises. A secondary aim was to analyse the muscle ratios between the vastus medialis oblique and vastus lateralis, along with the hamstrings to quadriceps ratio and the gluteus maximus to biceps femoris ratio.

Objectives: Electromyography (EMG) was used to monitor the activity of the hip and knee muscles during twenty rehabilitation exercises. The normalised data was used to identify a continuum of exercises, based on the extent to which each muscle was activated. The muscle ratios were also calculated, allowing the identification of a scale of exercises to preferentially activate certain muscles.

Subjects: Eighteen physically active volunteers participated in the study (males: n = 9, females: n = 9, mean ± standard deviation, age: 20 ± 1.3 years; height: 168.1 ± 9.7 cm; mass: 64.1 ± 9.8 kg).

Method: Surface EMG was used to measure the muscle activity of the gluteus maximus, gluteus medius, biceps femoris, rectus femoris, vastus medialis oblique and vastus lateralis during exercises which ranged between a straight leg raise and a weighted squat. The exercises were performed in a randomised order and three trials were performed of each. The muscle activity was normalised to a maximal voluntary isometric contraction specific for each muscle. The muscle ratios were calculated using specific equations.

Results: The counter movement jump and single-leg vertical jump frequently resulted in the production of the greatest EMG activity for each of the muscles, whilst the mini squat produced minimal muscle activity across all of the muscles. The bridging exercises activated the quadriceps to the least extent, resulting in these exercises producing the greatest hamstrings to quadriceps ratio. For the vasti ratio, the single-leg squat to 60° of knee flexion produced the greatest results. The step up exercise produced the highest gluteus maximus to biceps femoris ratio.

Conclusions: The continuum of exercises was identified for the activity of each muscle in order to aid clinicians by providing a guide from non weight bearing exercises through to functional jumps. This will ensure exercises are performed at the correct stage of rehabilitation to continually bring about muscular adaptations.

Contents

List of Figures

List of Tables

Abbreviations

EMG	Electromyography
ACL	Anterior Cruciate Ligament
GMax	Gluteus Maximus
GMed	Gluteus Medius
BF	Biceps Femoris
RF	Rectus Femoris
VMO	Vastus Medialis Oblique
VL	Vastus Lateralis
MVIC	Maximal Voluntary Isometric Contraction
PFPS	Patellofemoral Pain Syndrome
HS:Quads	Hamstring to Quadriceps Ratio
VMO:VL	Vastus Medialis Oblique to Vastus Lateralis Ratio
Q Angle	Quadriceps Angle
ICC	Intraclass Correlation Coefficients
SEM	Standard Error of Measurement
SENIAM	Surface Electromyography for the Non-Invasive Assessment of Muscles
IZ	Innervation Zone
CKC	Closed Kinetic Chain
OKC	Open Kinetic Chain
GMax:BF	Gluteus Maximus to Biceps Femoris Ratio

Chapter One: Introduction

1.1 Rationale

It has been estimated that 15.3 million people above the age of 16 participate in regular sporting activity each week throughout England (Sport England, 2013). Injuries to the lower extremity account for over half of all injuries across a range of sports, from football to gymnastics (Hootman et al., 2007). Specifically, the incidence of knee injuries from sport account for between 15-50%, depending on the type of activity (de Loes et al., 2000). Although hip injuries are less prevalent than injuries to the knee, certain sports participants are particularly at risk, including runners (van Gent et al., 2007). Muscle atrophy is often experienced as a result of an injury and can have prolonged effects, including joint instability (Andersen et al., 2006) and the pre-disposition to overuse conditions (Hollman et al., 2006). Abnormal hip mechanics have been identified to alter the valgus forces at the knee (Powers, 2010) and increase the risk of injuries such as anterior cruciate ligament (ACL) sprains (de Marche Baldon et al., 2009). Therefore, as part of the treatment for musculoskeletal injuries, there is a high focus on performing rehabilitation exercises in order to return the athlete to pre-injury levels of strength and function, and as a preventative method to stop injuries occurring in the first place (Heiderscheit et al., 2010). However, it was suggested that symptoms can deteriorate if the rehabilitation is inadequate at targeting the required muscles (Bolgla et al., 2008).

Due to these reasons, previous literature has investigated the validity of numerous exercises, which are commonly prescribed by clinicians after musculoskeletal injuries, such as squats (Andersen et al., 2006), hip hitches (Bolgla & Uhl, 2005) and step ups (Mercer et al., 2009). Electromyography (EMG) was used within these studies to quantify the amount of EMG muscle activity occurring during the exercises. However, one limitation which was often observed within the literature was that a restricted number of muscles were examined; for example, the research either focussed on gluteal activation (McBeth et al., 2012) or the vasti muscles (Bolgla et al., 2008). At present, there is an inadequate amount of evidence which incorporates both the hip and knee musculature to view the relationships within the lower limb kinetic chain, an aspect which is crucial for sports performers as the muscles have to work in unison (McMullen & Uhl, 2000). Also, an insufficient amount of research currently exists which incorporates the whole continuum of rehabilitation exercises from non weight bearing movements through to fully functional exercises with the use of an additional external resistance. It is essential that rehabilitation is progressive because the exercise stimuli need to continually increase and challenge the muscles to produce muscular adaptations (Murton & Greenhaff, 2010).

The exercises, which were analysed in the present study, are ones commonly used by practitioners including: straight leg raise, side-lying hip abduction, supine bridge, squat with a weighted bar and single-leg squat. Surface EMG was used to measure the activity of the gluteus maximus (GMax), gluteus medius (GMed), biceps femoris (BF), rectus femoris (RF), vastus medialis oblique (VMO) and vastus lateralis (VL). The EMG activity was expressed as a percentage of a maximum voluntary isometric contraction (MVIC) for each muscle and therefore the exercises could be displayed on a continuum depending on the level of muscle activity. The ranked order of exercises suggest a suitable progression during rehabilitation.

The GMax and GMed EMG activity levels are especially important due to their effect on lower limb biomechanics (Reiman et al., 2009). Weaknesses in these muscles have been associated to an increase in femoral anteversion (Nyland et al., 2004). Femoral anteversion is defined as the angle in the axial plane, by which the femoral neck deviates from the posterior femoral condyles; the anterior rotation of the femoral neck is also accounted for (Kendoff et al., 2007; Tayton, 2007). Excessive internal rotation of the tibia and over pronation at the subtalar joint are additional consequences of this biomechanical alteration (Hollman et al., 2006). These have been associated with an increased likelihood of overuse injuries such as iliotibial band syndrome (Ferber et al., 2010) and stress fractures of the distal tibia and tarsals (Tomaro et al., 1996).

The current study also analysed the ratios between several muscle groups, so the findings can be applied to the rehabilitation of certain injuries, which require specific muscles to be strengthened, including Patellofemoral Pain Syndrome (PFPS), post knee surgery, ACL sprains and knee osteoarthritis. There have been numerous factors associated with the development of PFPS including a weakened VMO and a delay in the onset of this muscle (Cowan et al., 2001). During rehabilitation for PFPS, the VMO:VL ratio therefore needs to be high in an attempt to preferentially activate the VMO (Balogun et al., 2010). Williams (2012) previously investigated the effects of hip positioning and knee range of motion on the VMO:VL ratio during wall-slide exercises. The analysis of additional exercises in the current study can only benefit the clinicians’ knowledge on treatment for PFPS. This work can also be applied to the rehabilitation of patients post surgery, as the VMO was shown to be the first muscle to show signs of atrophy after knee surgery (Sakai et al., 2000); thus athletes need to specifically re-strengthen this muscle.

The hamstrings to quadriceps (HS:Quads) ratio was another area for research within this study. Injuries to the ACL are prevalent, accounting for up to 80% of all knee ligament injuries (Gianotti et al., 2009). The ACL prevents anterior shearing forces on the tibia so when trying to protect this structure, the hamstrings should be strengthened because they also resist anterior translation of the tibia on the femur, which is caused by the force from the activation of the quadriceps (Chappell et al., 2007). Therefore, it is imperative that investigations are performed to identify the most effective rehabilitation and prehabilitation exercises to increase the HS:Quads ratio (Shields et al., 2005). It has also been identified that patients with knee osteoarthritis have an altered HS:Quads ratio, with the hamstrings activating to a larger extent than the quadriceps (Hortobágyi et al., 2005). This imbalance needs to be rectified during rehabilitation by specifically strengthening the quadriceps.

The GMax and BF should activate as synergists in a force couple relationship to extend the hip (Wagner et al., 2010), but it was identified that a weakness in one of the muscles may lead to an impairment of the other (Jonkers et al., 2003). If the GMax strength is not optimal, the contribution of the hamstrings may increase (Chance-Larsen et al., 2010; Lewis et al., 2007). To ensure correct gait and function (Lyons et al., 1993), it is therefore necessary to make certain these muscles are balanced, which could mean rehabilitation exercises are required to specifically strengthen one of the synergists. The GMax:BF ratio results within this study will provide clinicians with a continuum of exercises which either favour the activity of the BF or the GMax. This can be utilised to adapt the choice of rehabilitation exercises for the specific needs of individuals.

1.2 Aims

The aim of this study was to use EMG to quantify hip and knee muscle activity during twenty lower limb rehabilitation exercises, in order to identify a continuum of exercises in relation to the amount of muscle activation for the GMax, GMed, BF, RF, VMO and VL. A secondary aim was to analyse the VMO:VL ratio, HS:Quads ratio and GMax:BF ratio during the same exercises.

1.3 Objectives

- Measure the EMG activity of the hip and knee musculature, including the GMax, GMed, BF, RF, VMO and VL, during twenty lower limb rehabilitation exercises.
- Normalise the EMG values in order to identify a continuum of exercises in relation to the extent of muscle activity, expressed as a percentage of the MVIC.
- Measure the knee flexion angles during the jumps and step-based exercises.
- Calculate the VMO:VL ratios, HS:Quads ratio and GMax:BF ratios for each exercise.
- Identify a progressive order of exercises in relation to the results from the three muscle ratio calculations. This determines which exercises specifically activate one muscle to a greater extent than a different muscle.

Chapter Two: Review of Literature

2.1 Stages of Rehabilitation

2.1.1 Introduction

For the purpose of this study, the twenty exercises were categorised into stages, respective of the amount of body mass being supported by the dominant limb. However, it is important for the reader to be aware that these are not rigid stages of rehabilitation; the exercises link together in a continuous flow and some are, in fact, performed concurrently depending on the individual situation of the injury. This is the reason that the mean EMG activity for this study was shown on a continuum rather than specifically assigned to certain categories of high, medium and low percentages of MVIC.

There are no universally-recognised stages for rehabilitation; each study has their own interpretation and definition (Taylor & Taylor, 1997). Mithoefer et al. (2012) used a three phase system comprising of 'joint activation', 'progressive loading with functional joint restoration', and 'activity restoration'. Within the study, the first phase was predominantly partial weight bearing, range of motion exercises, phase two progressed to be full weight bearing and the final stage was sport specific reconditioning. Throughout the literature, the underlying trends appear similar for classifying the stages but Mattacola et al. (2002) instead labelled the stages as 'early', 'intermediate', 'advanced', and 'activity-specific'. A more detailed criteria was used in a study for post-operative rehabilitation of ACL ruptures; progression through the levels was only allowed if certain requirements were met, for example, having the ability to perform a single limb isometric squat and a hop for distance within 15% of the uninjured leg (Myer, Paterno et al., 2006).

Partial weight bearing exercises are often performed to protect and unload the injured area, when full weight bearing exercises could endanger the injured site (Vasarhelyi et al., 2006). Throughout this study, partial weight bearing was defined as having anywhere between 1-99% of body mass distributed through the dominant limb, whilst full weight bearing was when the dominant limb supported 100% of the body mass. Gait re-education is performed concurrently with rehabilitation exercises and is a clear way to display the progression of weight bearing stages; the patient begins using crutches with no weight supported on the injured leg, then progresses to partial weight bearing whilst still using the crutches, and finally unassisted walking (Starkey, 2013). The intensity of the exercises must reflect this. After reviewing the literature, the following criteria will be used to define each stage of rehabilitation during this study (Table 1).

Table 1. The criteria for the four stages of exercises.

Stage	Criteria	Criteria for progression	References
Early	• Non weight bearing • To improve range of motion	• Ability to partial weight bear • Full passive range of motion • Minimal or no effusion • Minimal or no pain	• Mattacola et al. (2002) • Starkey (2013)
Intermediate	• Partial weight bearing • To improve strength	• Ability to fully weight bear • Full active range of motion • Ready to return to running	• Mattacola et al. (2002) • Vasarhelyi et al. (2006)
Late	• Full weight bearing • To improve function • To further improve strength	• All sports specific exercises pain-free • Regained pre-injury levels of sports skills	• Mithoefer et al. (2012) • Myer, Paterno et al. (2006) • Vasarhelyi et al. (2006)
Performance Enhancing	• Full function • Totally recovered from injury	• Restoration of single leg power • Between 90-100% strength/power compared to uninvolved limb	• Lorenz and Beauchamp (2013) • Myer et al. (2011)

2.1.2 Rehabilitation versus Performance Enhancing

It needs to be noted that rehabilitation exercises and performance enhancing exercises are not two separate entities (Lehman, 2006); their effects can overlap by providing the athlete with increased strength and improved biomechanics, which both aid the recovery of an injury and develop sporting performance (Askling et al., 2003; Myer, Paterno et al., 2006). When referring to enhanced biomechanics, one aspect of this relates to core and gluteal strength, providing pelvic stability; this reduces lower back pain as well as increasing the efficiency of the force transfer between the upper and lower limbs to help with sporting technique (Kibler et al., 2006). This is the reason that within this study, both rehabilitation exercises and perceived performance enhancing exercises were investigated on the continuum. Resistance exercises have become an integral part of athletes' training to improve performance by increasing the muscles' force-generating capabilities, but it is still important that the exercises replicate, to some extent, the sports specific movements and contraction velocity (Young, 2006).

Squats are prescribed within rehabilitation but can also be used for performance purposes, as shown in a study by Wisløff et al. (2004). It was identified that in the seventeen elite football players who were tested, maximal squat strength showed a high correlation to 0-30 m sprint performance as well as vertical jump height. Furthermore, Nordic curls have already been identified as an effective exercise during injury prevention for hamstring strains, as well as during late stage rehabilitation and to enhance performance (Askling et al., 2003; Mjølsnes et al., 2004). Both of these studies focussed on a 10 week hamstring programme for footballers; Mjølsnes et al. (2004) concluded that Nordic curls increased eccentric hamstring strength to a significantly greater extent than hamstring curls, whereas Askling et al. (2003) focussed on the injury incidence during the season after completing such an exercise. At a minimum, exercises that quickly return athletes to full fitness after injury have a positive effect on performance by allowing an increased time for training (Hibbs et al., 2008).

Plyometric exercises incorporate an eccentric phase followed by a rapid concentric contraction, which stimulates the stretch-reflex (Impellizzeri et al., 2008). This type of exercise has been found to benefit female athletes by stabilising the knee and correcting biomechanics associated with ACL injuries, such as a large knee valgus force (Chimera et al., 2004). Miller et al. (2006) focused on the exercises in relation to performance and identified six weeks of plyometrics had the ability to improve agility skills, especially important to footballers and many other sports players (Steffen et al., 2008). The counter movement jump and single-leg vertical jump were incorporated into the current study to determine the amount of muscular activity and co-contraction levels compared to more controlled, less explosive closed kinetic chain (CKC) exercises, such as the deadlift, squat and bridge exercises. The term CKC is defined as a multi-joint movement with the distal segment in contact with a fixed surface, whereas the phrase open kinetic chain (OKC) is defined as a non weight bearing, single-joint movement (Irish et al., 2010). The quadriceps setting exercise and straight leg raise are two examples of OKC exercises used within this study.

2.2 The Use of Electromyography

2.2.1 Introduction

Electromyography is commonly used to research muscle function by assessing the myoelectric signals (Dowling, 1997). Voluntary muscle contractions are controlled by the central nervous system sending signals to the alpha-motor anterior horn, which proceed to modify the diffusion characteristics of the muscle fibre membrane, allowing sodium (Na+) ions to flow in (Kleissen et al., 1998; Konrad, 2005). When a specific threshold is exceeded during the Na+ influx, it causes an action potential to occur (Roatta & Farina, 2010). The EMG signal represents the action potentials, which are due to the depolarisation and repolarisation from the exchange of ions through the semi-permeable muscle cell membrane (Kleissen et al., 1998). Within the detectable range of the electrodes, all motor unit action potentials are superposed to produce a raw EMG signal; an equal distribution of negative and positive amplitudes are observed, thus the mean values equal zero (Konrad, 2005).

Surface and intra-muscular EMG are two types of electrodes that can be used. The intra-muscular electrodes analyse specific motor units and are thought to be less prone to cross-talk (Dowling, 1997), whereas the surface electrodes examine a larger superficial area but are less intrusive (Kleissen et al., 1998; Soderberg & Knutson, 2000). Chapman et al. (2010) concluded similar global muscle recruitment patterns when using surface EMG and fine-wire intra-muscular EMG during a cycling task.

In order to obtain an effective EMG signal, the user needs to be aware of the possible sources that can influence the data (Soderberg & Knutson, 2000). De Luca (1997) identified intrinsic factors such as the number of active motor units, fibre diameter and subcutaneous tissues, and extrinsic causative factors including the electrode configuration, which incorporates the shape and size, as well as the electrode orientation in relation to the muscle fibres. The Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) guidelines are often utilised to reduce these risks. The SENIAM project is a peer-reviewed European project which has investigated sensor placement and procedures (SENIAM, 2013). The SENIAM guidelines suggest the use of bipolar silver/silver chloride 10mm electrodes, with an inter-electrode distance of 20 mm.

2.2.2 The Reliability of Electromyography

'Reliability' is defined as the level to which measurements are repeatable and consistent, whereas the term 'validity' refers to the meaningfulness of the research (Drost, 2011). The most important factor to identify, when investigating the reliability of EMG activity, is whether a real change has occurred; for example, has the subject become stronger?, or are the discrepancies due to a systematic or random bias? (Olds, 2002). The systematic bias is shown as a constant difference for all subjects in the same direction, which can be as a result of factors including fatigue or the learning effect (Atkinson & Nevill, 1998; Olds, 2002). Studies have been performed to assess the reliability of EMG between subjects and sessions. The intraclass correlation coefficients (ICC) for quadriceps MVICs have been as high as 0.99 (McCarthy et al., 2008) with a low standard error of measurement (SEM) of 1.1-6.4% of the mean, showing high repeatability (Rainoldi et al., 2001). Although these figures appear encouraging, assessing the reliability of EMG is a complex process in terms of which methods are most valid and which equations should be used (Atkinson & Nevill, 1998). There are numerous methods for calculating the ICC, depending on the inclusion or exclusion of the systematic error and whether it is a 1 or 2 way model (Weir, 2006). Rousson et al. (2002) suggested the ICC should be used for testing the inter and intra-rater reliability whereas the product moment coefficient should be used to examine the reliability between repetitions for subjects. The negative point of the product moment coefficient is the lack of consideration for systematic bias (Atkinson & Nevill, 1998), therefore its use has been discouraged (Weir, 2006).

2.2.3 Electrode Placement

Due to differing aims and objectives within previous studies, authors have chosen to investigate different muscles. There is no consensus within the literature as to the ideal electrode placement locations but those commonly used have been reported. Studies by Ayotte et al. (2007) and Escamilla et al. (2010) gave precise locations for the positioning of the electrodes in relation to bony landmarks. The SENIAM guidelines provide similar instructions for certain muscles including the RF, BF and GMax but not the hip adductors. However, other literature used vague instructions such as 'over the muscle belly', in a parallel arrangement (Boudreau et al., 2009; Farrokhi et al., 2008). Rainoldi et al. (2004) researched the consistency of innervation zones (IZ) between individuals to determine the optimal positions for surface EMG electrodes. The IZ is the location where the motor unit action potentials originate and propagate along the muscle fibres (Saitou et al., 2000). The results showed that for some muscles, electrodes could be positioned in relation to the bony landmarks, due to high uniformity of the IZ location, such as with the BF and VL; however, for other muscles, the IZ needs to be found for optimum electrode placement (Rainoldi et al., 2004).

Some studies tested one hamstring muscle and generalised the values to the whole hamstrings muscle group (Ayotte et al., 2007; Farrokhi et al., 2008), as it was assumed that the three muscles all had equal ratios for performing knee flexion (Kellis et al., 2012). The hamstrings muscle activity was measured using EMG and it was identified that the BF produced more reliable EMG activity than the semitendinosus (Kellis & Katis, 2008), hence some studies have only recorded the activity of this lateral hamstring within their testing (Ayotte et al., 2007; Farrokhi et al., 2008; Ryu et al., 2011). Foot position must be neutral though, because it was concluded that the medial to lateral hamstring activation ratios differed with foot rotation (Lynn & Costigan, 2009).

When considering the quadriceps, Ayotte et al. (2007) only recorded the EMG activity for the VMO so the data analysis was based on the assumption that this muscle activity reflected the work of the other vasti muscles and the RF. This should be viewed with caution because the VMO only extends the knee, whereas the RF activates over the knee and hip (Hagio et al., 2012). Furthermore, the VMO was found to be innervated by a different nerve supply to the rest of the vastus medialis, increasing the likelihood that its activation was not a true replication of the other quadriceps (Toumi et al., 2007). In addition to this, Rainoldi et al. (2001) attributed the relatively smaller size of the VMO to the increased difficulty in positioning the electrodes in the same place between days and subjects, compared with the VL so extra care was taken within the current study.

During dynamic exercises, the surface EMG electrode can fractionally displace in relation to the underlying muscle (Ekstrom et al., 2012). This can affect the intensity of the EMG signal and can alter the wave form (Massó et al., 2010) so it is an important factor to consider. However, researchers have suggested ensuring there is strain relief on the connecting wires to decrease this risk and therefore overall high reliability can still be obtained (Fauth et al., 2010).

When completing testing sessions over various days, some authors have used acetate sheets to mark where electrodes should be positioned, whereas others have used permanent pen to directly draw on the skin (McCarthy et al., 2008; Rainoldi et al., 2001). During the review of literature, there were no studies found, which compared these methods. Ensuring the environmental conditions are the same day by day is important, as variations in skin temperature, as well as the participants' mood and motivation, can all affect the reliability of the EMG activity (Rainoldi et al., 2001). The amount of verbal encouragement given can also have an effect; in order to produce a maximal force, encouragement may need to be given (McNair et al., 1996). Taking all the above into consideration, Kellis and Katis (2008) investigated the test-retest reliability of the hamstrings and concluded the ICC was 0.41-0.96 during ramp isometric contractions for the BF, which the authors classed as between poor and high reliability. However, there is no consensus within the literature with regards to the scoring system so in the future, a recognised system should be developed to suggest what constitutes poor, moderate and high reliability results (Weir, 2006). Kellis and Katis (2008) identified an increase in the ICC when the subject had to produce less force, represented by lower EMG activity. This opposed the results of Ng et al. (2003), who found a higher reliability at maximal efforts compared to sub maximal trials.

2.3 The Normalisation Process

2.3.1 The Use of Maximal Voluntary Isometric Contractions

Due to the natural variance and factors affecting the collection of EMG activity, the need for a normalisation process has been widely recognised, especially when comparing between subjects and days (Ball & Scurr, 2010). A reference task is completed so the EMG activity during an exercise can be expressed as a percentage, relative to the reference value. One study concluded that for the quadriceps, MVICs produced significantly less EMG activity compared to a maximum concentric contraction, therefore, it was recommended to use dynamic reference tasks (Ekstrom et al., 2012). However, MVICs are commonly used, thus making it easier to compare between studies (Bolgla & Uhl, 2005; Lubahn et al., 2011; O'Sullivan et al., 2010). This method of normalisation has limitations though, including the assumption that the subject performed the exercise to their highest ability and whether the task was the most appropriate one to target the required muscle (Rainoldi et al., 2001).

Previous studies have recorded the exercise EMG activity to be superior to that of the MVIC, shown by a figure in excess of 100% MVIC (Boren et al., 2011; Jacobs et al., 2009). Although this data was able to highlight trends within each of the studies, it made the comparison between the literature to be unfeasible. The values in excess of 100% MVIC could have been due to the supposed MVIC not actually being maximal due to diminished motivation levels or the patient positioning. When performing MVICs, some studies have used manual muscle tests with the subject contracting against the examiner's resistance (Lubahn et al., 2011; McBeth et al., 2012; Park et al., 2010). This may have led to diminished effort levels depending on the strength of the examiner and the stability of the resistance (Shenoy et al., 2011; Silvers & Dolny, 2011). Other studies have used an isokinetic dynamometer (Ayotte et al., 2007; Matheson et al., 2001; O'Sullivan et al., 2010). Shenoy et al. (2011) identified significant differences in MVICs for the quadriceps and hamstrings when using manual muscle testing as opposed to an isokinetic dynamometer. However, Lin et al. (2008) compared the quadriceps femoris EMG activity for manual muscle testing and a Cybex isometric dynamometer; no statistical differences were found between the EMG activity when using the two methods. Methodological differences could have led to these contrasting results. For example, the angle of the knee differed between studies and also Shenoy et al. (2011) only recorded data for one trial per muscle group, so for each subject, anomalies were unable to be identified as there were no other values to compare the results to.

The use of a resistance belt has also become a common practice for measuring MVICs as it acts as a compromise between manual muscle testing and expensive dynamometer equipment (Boren et al., 2011; Nyland et al., 2004). There is limited evidence to show the reliability of this method, but it should be viewed favourably because it addresses some of the issues highlighted before with the manual testing, including: it is a fixed resistance, the strap is firmly secured providing stability, and it does not rely on human strength (Silvers & Dolny, 2011).

2.3.2 Positioning for Maximal Voluntary Isometric Contractions

Another factor to acknowledge when obtaining an MVIC is the range of motion at which the data is collected. For example, with the GMed, many studies performed side-lying hip abduction (Ayotte et al., 2007; Bolgla & Uhl, 2007; Ekstrom et al., 2007, McBeth et al., 2012; Park et al., 2010). However, the position of the leg differed between being in 0° of abduction (Ayotte et al., 2007), 25 to 35° (Bolgla & Uhl, 2005; Bolgla & Uhl, 2007; McBeth et al., 2012; Park et al., 2010) or end of range abduction (Ekstrom et al., 2007). In the past, studies have identified muscles to be strongest in mid-range, so testing abduction at the end of range could immediately have led to less EMG activity (Folland et al., 2005). The length of the muscle affects the EMG amplitudes due to the increase in motor unit discharge when the muscle is shorter, and from the alteration in neural activation (Desbrosses et al., 2006). The muscle length and fibre diameter also influence the conduction velocity; the shorter the muscle, the greater the velocity (Masuda & De Luca, 1991).

Other studies have measured the GMed MVIC with the subject standing and completing hip abduction (Dywer et al., 2010; O'Sullivan et al., 2010). Discretions within the results could be evident because this non weight bearing, standing movement was identified to produce less EMG activity of the GMed than weight bearing or side-lying abduction; thus not representing 100% of the muscle's capability (Bolgla & Uhl, 2005). Some studies have lacked control during the collection of data for the MVIC. Ekstrom et al. (2007) and McBeth et al. (2012) tested the GMed in a side-lying position with the hip in "slight" extension, whilst a different study positioned the subject in "slight" hip abduction (Jacobs et al., 2009). These are vague instructions, lowering the repeatability of the movement, so these results should be viewed with caution.

Considering the testing angle is also relevant when recording the EMG activity for the quadriceps. Lin et al. (2008) measured when the knee was at a 45° angle, whereas other studies have utilised a 60° angle (Ayotte et al., 2007; Farrokhi et al., 2008; Matheson et al., 2001). Some studies have even failed to specify a range (Dwyer et al., 2010) . Forty-five degrees was chosen by Lin et al. (2008) according to the work by Burden (2003), even though the 60° angle has been identified to produce the greatest isometric contraction (Matheson et al., 2001).

For the hamstrings, there have been larger adaptations for the MVIC testing positions throughout the current literature. These have varied between: lying prone with 45° knee flexion (Ekstrom et al., 2007), lying prone with combined knee flexion and hip extension (Cambridge et al., 2012) and finally, lying supine with the hip and knee flexed to 90° whilst simultaneously flexing the knee further against resistance from a 45 cm stool (Farrokhi et al., 2008).

With respect to the MVIC for the GMax, the subjects have performed hip extension whilst having the knee flexed to 90° (Lubahn et al., 2011). However, differences were evident as to whether the subject completed this in a prone position (Ekstrom et al., 2007; Farrokhi et al., 2008) or whilst standing (Dwyer et al., 2010). Gravity and the stability of the pelvis may have affected the maximum force produced during the standing position (Jung et al., 2012).

2.3.3 Verbal Encouragement

Between studies, there is no standardised procedure for verbal encouragement. Boudreau et al. (2009) did not give any encouragement, whereas many other studies have done so (Bolgla & Uhl, 2005; Jacobs et al., 2009; Lin et al., 2008; Matheson et al., 2001; O'Sullivan et al., 2010). The amount of encouragement and volume could have impacted the EMG activity (Boren et al., 2011). McNair et al. (1996) found maximal contractions to be 5% higher when verbal encouragement was given, which is a desired outcome during an MVIC. Therefore, within the current study, the same examiner will always provide verbal encouragement during the MVIC.

2.3.4 Data Analysis of the Maximal Voluntary Isometric Contractions

As mentioned before, the MVICs in the study by O'Sullivan et al. (2010) were possibly submaximal, either from the subject positioning, motivation levels or because the data analysis encompassed the full duration of the 5 second contraction. Other studies have attempted to utilise the greatest EMG activity by only analysing a section of the MVIC contraction; for example, the middle three seconds (Lin et al., 2008), the middle 1.5 seconds (Ayotte et al., 2007) or the highest one second interval (Escamilla et al., 2010), which may produce more valid results. There were also differences between whether the mean (Boudreau et al., 2009; Dwyer et al., 2010) or peak EMG values were used (Andersen et al., 2006; McBeth et al., 2012).

When calculating the root mean square (RMS), large time intervals may have affected the EMG activity by excessively flattening the data (Wong, 2009). Within the reviewed literature, the RMS time intervals differed between 15 ms and 500 ms which would have led to discrepancies when comparing between the studies (Bolgla & Uhl, 2007; Jacobs et al., 2009). The most appropriate RMS time frame interval cannot be stated as it can be dependent upon the EMG equipment and analysis software. Only a few studies performed the MVIC before and after the exercise testing to ensure the reliability was high (Andersen et al., 2006; Bolgla & Uhl, 2005; Bolgla & Uhl, 2007). Matheson et al. (2001) actually discarded the data if the pre and post MVICs were not within two standard deviations but this meant they had to recruit a larger amount of subjects as only 16 sets of data were analysed out of the 52 subjects who volunteered for the study.

2.4 The Use of an Electrogoniometer

An electrogoniometer is regularly used to monitor the knee flexion angles (Andersen et al., 2006; Jacobs et al., 2009; McBeth et al., 2012). Again, there are no set guidelines but the positioning used by Brandon et al. (2011) was described as one arm of the electrogoniometer being along the line between the lateral femoral condyle and greater trochanter, and the other arm between the head of fibula and lateral malleolus. When the knee flexes, the centre of the knee joint rotation changes due to the femoral condyles translating anteriorly or posteriorly depending on whether the knee is loaded (Hill et al., 2000), making it difficult to track the knee angle with a hand held goniometer (Tesio et al., 1995). This is a positive aspect of electrogoniometers, because they utilise the relative position of the femur and the shank to calculate the knee flexion angle rather than having to rely on being placed over the centre of rotation (Piriyaprasarth et al., 2008).

Electrogoniometers were found to produce reliable results, with the mean coefficient of variation being 5.3% during a barbell squat (Brandon et al., 2011). Reliable angle results were also observed during a counter movement jump (Petushek et al., 2012). However, the same study found differences between the angles measured using the electrogoniometer and those from a two dimensional video analysis system, potentially due to video perspective errors. The intra-tester reliability (ICC) during a seated position, supine and standing, ranged between 0.75-0.88, with the error of measurement being less than 1.7° (Piriyaprasarth et al., 2008). These results were superior to those of the inter-tester reliability (ICC between 0.57-0.80). This highlights that when using the same examiner, they were more consistently able to re-position the electrogoniometer. However, van der Linden et al. (2008) only used one tester to measure joint range of motion and found poor repeatability measures when recording over two testing sessions. This may have been due to the length of time between the two assessments being as high as 31 days in some cases so a physiological change in the participants' flexibility may have occurred. Thus, these results should be viewed with caution. In the current study, one tester will therefore be used during the data collection process.

2.5 Subjects

Within the literature, the varied subject characteristics may be a reason for differences in the results. Age, gender, past medical history and level of sporting ability are all contributing factors. Frequently, the mean age of subjects ranged between 20 and 25 years old (Boudreau et al., 2009; Distefano et al., 2009; Dwyer et al., 2010; Matheson et al., 2001). However, not all studies followed this trend with some averages being in the 30s (Ayotte et al., 2007), 50s (Jacobs et al., 2009) or even the 70s (Mercer et al., 2009). The overall range of years may have also contributed to varied results; Ekstrom et al. (2007) included subjects from 19 to 58 years and Jakobsen et al. (2013) involved participants between 26 and 67 years. This non-specific subject pool could have made a large impact on EMG activity due to age-related changes affecting the neural control and muscle recruitment (Hortobágyi et al., 2011).

It is still unclear whether gender differences exist for the activation of lower limb muscles during functional tasks and exercises. One study found gender had no effect on the GMed, BF and RF muscle activation during a step down, forward lunge and side-step lunge (Bouillon et al., 2012). These findings were similar to that of Cowan and Crossley (2009), who found no differences in the EMG activation of the GMed, VMO or VL during a stair-stepping task. However, Dwyer et al. (2010) concluded females had higher RF and GMax mean EMG values during exercises such as a single-leg squat, lunge and step up and over. These contrasting results may have been due to inconsistent standardisation methods; Bouillon et al. (2012) used lunge distances and step heights which were relative to the height of the subject, rather than absolute values. However, Dwyer et al. (2010) only normalised the lunge distance to leg length but used standard step-heights and did not control the single-leg squat depth at all. Caterisano et al. (2002) and Wright et al. (1999) only used male subjects so it is important for future research to be more conclusive as to whether gender affects muscle activation, which would determine if the findings from these studies could be generalised to the wider population. This will be discussed further in Section 2.9 (page 40).

Specifically for the performance enhancing exercises, the subject's training history could have been a determining factor for the EMG activity due to the learning effect and amount of experience potentially altering technique (Brandon et al., 2011). Chapman et al. (2008) concluded that experienced cyclists had a more refined muscle recruitment pattern from neuromuscular adaptations as a result of repeated performances. It can be speculated that this also occurs in other sports and training exercises. Caterisano et al. (2002) used subjects with an extensive training history, greater than 5 years. During this time, the subjects had included weighted squats as a regular exercise in their training programme. A different study used bodybuilders and Division 1 American footballers, who had each completed the exercises in their training routine 3-12 times per month, for the previous three years (Wright et al., 1999). Nuzzo et al. (2008) used recreational athletes, instead. The recreational athletes would most likely reflect the average population throughout the United Kingdom so this calibre of subject was used within the current study.

Regardless of age, gender or training status, the medical history of the subjects needs to be considered. Various studies used healthy, asymptomatic subjects to record the muscle activity during rehabilitation exercises (Ayotte et al., 2007; Boren et al., 2011). However, it can be speculated that EMG activity may differ from subjects with an injury, as shown in the study by Cowan et al. (2002) comparing patients with PFPS to healthy controls. Alterations in neuromuscular control in injured subjects could have been the reason for the two groups of subjects responding differently (Chmielewski et al., 2006). Even when using symptomatic subjects, it was difficult for the studies to include a tight inclusion criteria, so not all subjects experienced identical symptoms for the same duration. For example, Callaghan et al. (2009) recruited subjects who had experienced peripatellar pain for longer than 6 months, with pain during at least one of the following: prolonged sitting, climbing or descending stairs, running, hopping, jumping or squatting. This was an extensive list, so the subjects were not representative of a specific patient group. Therefore, it is difficult to compare results between such studies with different subject inclusion criteria. When considering the type of subject, it is still important to investigate healthy participants in order to obtain a baseline for how the muscles should activate in such conditions, during specific exercises (Bolgla & Uhl, 2005). This was the purpose of the current study.

2.6 Rehabilitation Exercises

It is generally accepted that for strength gains to occur, the exercises must reach the 40% MVIC threshold, indicating the neuromuscular activation exceeds the limit for muscular adaptations (Ayotte et al., 2007; Escamilla et al., 2010). It appears that the greater the muscle activation, the greater the gains (Andersen et al., 2006). This concept will be referred to throughout the rest of the thesis.

2.6.1 Activating the Gluteus Maximus

The exercises which have been identified to activate the GMax to the greatest extent include: front plank with hip extension (106% MVIC), unilateral wall squat (86% MVIC) and gluteal squeeze (81% MVIC) (Ayotte et al., 2007; Boren et al., 2011). As explained previously, the EMG activity in the study by Boren et al. (2011) showed very high percentages compared with other studies, potentially from a submaximal reference contraction. Ayotte et al. (2007) concluded the unilateral wall squat produced greater EMG activity of the GMax than a unilateral mini-squat and forward, lateral or retro step up. This could have been due to the foot being placed more anterior to the hip and centre of mass during the wall squat, which could have affected the muscle activation (Blanpied, 1999). Boudreau et al. (2009) found a single-leg squat (35% MVIC) to produce the greatest GMax EMG activity compared to a lunge (22% MVIC) and step-up-and-over (17% MVIC), but the results were of a much lower percentage overall. During the normalisation process, it was not stated that the reference value was maximal, and the depth and speed of squats were not standardised which could have altered the EMG activity. During the exercises, the trunk position in relation to the base of support, as well as the direction of the lunge and step up, could have affected the EMG activity and needs to be considered by clinicians (Reiman et al., 2012). Trunk position is also important during bridging exercises, because it was identified that a posterior tilt of the pelvis significantly increased the GMax EMG activation (Ishida et al., 2011).

The GMax is the primary hip extensor and external rotator, so if this muscle has poor endurance capabilities, transverse plane movements may be less controlled (Lubahn et al., 2011). This relates to the findings of a study by Souza and Powers (2009) who concluded that women suffering from PFPS had decreased hip extension endurance, which was a good predictor of how much hip internal rotation occurred during the running gait. Therefore, the use of exercises, such as the front step up (34.4% MVIC), which only just failed to reach the 40% MVIC threshold for strength gains, can instead be used as a high repetition exercise to increase GMax endurance (Escamilla et al., 2010; Lubahn et al., 2011). Having said this, Ayotte et al. (2007) identified the step up to activate the GMax to a much larger degree of 74% MVIC. Differences for this could be due to the height of the step, although it was difficult to analyse this because Lubahn et al. (2011) did not specify the height. In a study by Ekstrom et al. (2012), it was suggested that where possible, a 30-40 cm step should be used to ensure the exercise is sufficient for strengthening the musculature.

2.6.2 Activating the Gluteus Medius

Back in the 1980's, a paper was published which identified the GMed as having three sections (anterior, middle and posterior) with their primary function being to stabilise the hip and pelvis (Gottschalk et al., 1989). More recently, it was concluded that the different portions activated to different levels during weight bearing single-limb exercises (O'Sullivan et al., 2010). For the highest GMed activation, the suggested exercises have been side plank with abduction of the non-dominant leg (103% MVIC), single-leg squat (82% MVIC) and side-lying abduction (81% MVIC) (Boren et al., 2011; Distefano et al., 2009).

Side-lying hip abduction has often produced high mean EMG activity but there have been variations in the method used between studies, including the hip position and location of the resistance, if any. During the side-lying abduction exercise, McBeth et al. (2012) tried to control the pelvic alignment by placing a pressure cuff under the anterior aspect of the pelvis to monitor any tilting. Park et al. (2010) investigated the differences between exercises with and without a pelvic compression belt; differences were evident in the GMed EMG activity, thus it can be hypothesised that simultaneous deep core muscle activation can affect the GMed EMG activation between subjects. If the subject unintentionally rotated the hip whilst abducting the hip, discrepancies may have been seen because it was identified that hip abduction with internal rotation was ranked higher at activating the GMed than hip abduction with external rotation (Philippon et al., 2011). McBeth et al. (2012) had hypothesised that side-lying hip abduction with external rotation would increase the EMG activity of the GMed and GMax as a hip external rotator, and decrease the contribution of the tensor fascia latae (TFL). However, this was not evident from the findings. Instead, the TFL and hip flexor activity increased which was potentially due to the subject rolling the body posteriorly, so gravity worked to produce a slight hip extension force, thus the anterior muscles had to contract to a greater extent to ensure the hip remained in a neutral position (McBeth et al., 2012).

When considering the torque around the fulcrum, suggestions can be made about the leg length of the subjects. To control for this, Jacobs et al. (2009) used trigonometry to calculate the appropriate resistance and height of abduction relative to the subject so the hip angle stayed consistent. Jacobs et al. (2009) highlighted that exercises should be specific to the patient and for those who were unable to adequately stand, side-lying exercises were sufficient at activating the GMed.

However, this muscle is predominantly used to stabilise the pelvis during weight bearing activities, hence exercises like single-leg squats and single-leg bridges are of great importance (Reiman et al., 2012). The significance of the weight bearing element was especially shown during a study by Youdas et al. (2013) who reported that during a resisted lateral-step, the standing leg had a significantly greater GMed EMG activation than the limb which was actively abducting the hip. Similar findings were concluded by Bolgla and Uhl (2005); the GMed on the weight bearing limb activated to 42% MVIC during a standing hip abduction task, whereas the non weight bearing leg only reached 33% MVIC.

With functional weight bearing exercises such as the single-leg squat, it is important to follow a standardised protocol, because it was identified that when subjects were told to intentionally co-activate the core muscles, the lower limb kinematics were altered which could affect the hip EMG activity (Shirey et al., 2012). A few studies have examined the mean EMG activity of the GMed during a pelvic drop, also known as a hip hitch, but the muscle activity varied between 24-52% MVIC (Bolgla & Uhl, 2005; O'Sullivan et al., 2010). It was hypothesised that the lower activity was as a result of the study using a more accurate MVIC value; three reference tasks were recorded (hip abduction, internal rotation and external rotation) but only the highest was used in the analysis (O'Sullivan et al., 2010). Differences in timings could also be responsible. Many studies used metronomes but the speed varied considerably (Bolgla & Uhl, 2005; Matheson et al., 2001).

The need for external resistance was questioned by Thorborg et al. (2010), who found no differences in hip abductor strength following two exercise programmes, which included a) body weight exercises or b) light external resistance. Also, Lubahn et al. (2011) utilised additional resistance from a cable machine to pull the knee medially during functional exercises, in a bid to increase the activation of the gluteals, by requiring the muscles to provide a greater external rotation torque and greater hip abduction torque. However, the EMG activity opposed this reasoning. Except for the double leg squat, the additional resistance actually decreased the gluteal EMG activity, potentially from initiating poor biomechanics by internally rotating the femurs.

2.6.3 Activating the Hamstrings

Ekstrom et al. (2007) tested nine rehabilitation exercises and only identified the single-leg bridge to activate the hamstrings above the 40% MVIC threshold for strength gains. Within the study, the BF was tested to represent the hamstrings.

When investigating hamstring activity, it is important to ensure the foot position and hip rotation is standardised because these factors can alter the activation of the medial or lateral hamstrings. It was concluded that external rotation preferentially activated the lateral hamstring, whilst internal rotation activated the medial hamstrings to a greater extent, during a bridge, hamstring curl and deadlift (Lynn & Costigan, 2009). The authors of the study proposed this could have been due to the hamstrings' role to control the transverse plane forces, including tibial rotation, or due to the pre-activation of the muscle during the rotation element, which increased the responsiveness of the particular hamstring during the exercise.

As well as differences in step lengths during a lunge exercise, the trunk position is an important factor to consider. It was concluded that a lunge with a forward lean produced significantly greater hamstring EMG activity than a lunge with an erect torso (17.9% MVIC and 11.9% MVIC respectively) (Farrokhi et al., 2008). This was likely to have been a result of the centre of mass shifting forwards during the leaning lunge (Riemann et al., 2012). Overall, hamstring activation during a lunge was generally low, with values not exceeding 18% MVIC (Ekstrom et al., 2007; Farrokhi et al., 2008), unless resistance from elastic tubing or dumbbells were added, but even then, figures did not reach the 40% MVIC threshold for strength adaptations (Jakobsen et al., 2013). Differing amounts of variance occurred between individuals within the studies, despite examiners attempting to reduce this by using a relative step length (Farrokhi et al., 2008). As the standard deviations appeared large for some results, some individuals may have benefitted from the exercises to a greater extent than others, depending upon their initial strength levels (Ekstrom et al., 2007).

From a simplistic mechanical view-point, the deadlift and squat may appear to be similar; however, this resemblance is minimal as they require different muscle activation and movement patterns (Hales et al., 2009). The deadlift exercise can be performed with either the bent leg or straight leg technique, both commonly used to strengthen the hamstrings (Nuzzo et al., 2008). This exercise has a high focus on the eccentric component of the hamstrings, especially important during the swing-through phase of sprinting (Askling et al., 2003). When analysed with EMG, the hamstrings produced a similar magnitude during the concentric phase of a straight leg deadlift and the concentric section of a hamstring curl. Both of these exercises produced double the EMG activation for the hamstrings than the back squat (Wright et al., 1999). Sumo deadlifts and conventional deadlifts were assessed in a study by Escamilla et al. (2002) but no differences were identified for the EMG activity of either the medial or lateral hamstrings during their hip extension role. During a 3-dimensional analysis of the two lifts, biomechanical differences were evident, but, with respect to the hamstrings, hip extensor moments were similar (Escamilla et al., 2000). The hamstrings activity was greatest during the 0-60° of knee flexion range of motion, with the GMax increasing during the final 30°, potentially to stabilise the deceleration of the lift as it reached the lowest point (Escamilla et al., 2002).

For squat based exercises, Caterisano et al. (2002) concluded that hamstring EMG activity was deemed low during all depths. During this movement, the hamstrings act as the antagonist for knee extension but as the agonist for hip extension (Wright et al., 1999). Because of their biarticular structure, their overall length remains similar throughout the movement, thus allowing a consistent level of force output, albeit low (Schoenfeld, 2010). The hamstrings also co-contract with the quadriceps to neutralise the anterior shearing forces on the tibiofemoral joint (Stuart et al., 1996).

2.6.4 Activating the Quadriceps

The quadriceps have been tested during a large range of rehabilitation exercises, though many studies have either focussed on just one muscle (Ayotte et al., 2007; Boudreau et al., 2009; Dwyer et al., 2010; Ekstrom et al., 2007), or the combination of the VMO and VL (Boling et al., 2006; Hertel et al., 2004; Wong et al., 2013). The RF is biarticular, crossing both the hip and knee joints, so may activate differently to the vasti muscles, which just extend the knee (Hagio et al., 2012). Hence, within the current study, the RF, VMO and VL were all assessed. The final muscle to complete the quadriceps femoris is the vastus intermedius, but this cannot be tested with surface EMG due to its' deep positioning beneath the RF (Waligora et al., 2009).

For a non weight bearing isometric quadriceps setting exercise, the VL was activated to 32% MVIC, whilst the RF produced a contraction to 24% MVIC (Andersen et al., 2006). During more functional weight bearing exercises, the VMO reached limits of 85% MVIC during a lateral step up (Ekstrom et al., 2007) and 66% MVIC during a unilateral wall squat (Ayotte et al., 2007). These both involved a single limb stance with flexion and extension of the knee. Many different types of step up exercises have been analysed with step height, direction and exercise protocols being the differing factors (Ayotte et al., 2007; Boudreau et al., 2009; Ekstrom et al., 2007). The lunge was another exercise to produce quadriceps values above the 40% MVIC threshold to improve strength, with the EMG activity in the literature varying between 45.6% MVIC and 76% MVIC (Ekstrom et al., 2007; Farrokhi et al., 2008). The differences may have been due to Ekstrom et al. (2007) measuring the EMG activation for the VMO, whereas Farrokhi et al. (2008) investigated just the VL.

2.6.5 Activation During a Squat

The squat exercise has been briefly referred to thus far. However, due to this study incorporating four different types of squats, more details will be provided. The squat is a commonly prescribed rehabilitation exercise, involving flexion and extension of the hips, knees and ankles (Schoenfeld, 2010). Maximal strength during such an exercise was found to be a determinant for sprint performance and jump height (Wisløff et al., 2004) so it was hypothesised this exercise can be used for injury rehabilitation and to enhance performance. Varied amounts of resistance can be applied to this exercise depending on the aim of the exercise session. Within the literature, this has ranged between using a 10-repetition maximum (Andersen et al., 2006), a percentage of a 1-repetition maximum (Hamlyn et al., 2007; Nuzzo et al., 2008) or a percentage of body mass (Baffa et al., 2012; Caterisano et al., 2002; Ninos et al., 1997). Brandon et al. (2011) used a different approach to reporting the amount of resistance by incorporating the mass of the bar plus 88% of the participants' body mass, as this represented the amount that had to be vertically displaced; the 12% shank and foot segments were excluded. There were also variations in how the resistance was applied. An Olympic bar appeared a popular choice (Hamlyn et al., 2007; Wright et al., 1999), whereas other studies used a weighted rucksack (Baffa et al., 2012; Ninos et al., 1997). This may have had a large effect on the amount of hip and knee muscle activity because it was suggested that by placing the Olympic bar above the centre of mass, as seen with a front or back squat, an unstable effect may have been apparent so the muscles were required to counterbalance this element (Hamlyn et al., 2007).

As expected, no standardised depth of squat was apparent when comparing all the research. One study categorised three types of squat: partial squat to 45° of knee flexion, parallel squat to 90^{o} of knee flexion and full squat to 135° of knee flexion (Caterisano et al., 2002). Whereas other studies grouped them into: partial squats (40°), half squats (70°-100°) and deep squats (> 100°) (Schoenfeld, 2010). For the purpose of these depths, 0° is defined as being when the knee is in full extension. This notation for knee flexion angles will be used throughout the rest of the thesis. When investigating the muscular activity using EMG, studies used squats to 70° (Brandon et al., 2011; Nuzzo et al., 2008), 100° (Andersen et al., 2006) or others instructed subjects to descend until their femur just dipped below horizontal (Wright et al., 1999). The depth of squats is an important factor to assess because it was found to affect the muscular activity, especially for the GMax, with the deeper squat producing greater EMG activity (Caterisano et al., 2002). For rehabilitation purposes, it needs to be remembered that the deeper the squat, the higher the patellofemoral compressive forces and greater the tension on the quadriceps and patellar tendons (Shoenfeld, 2010).

2.7 Data Analysis for the Exercise Trials

The data analysis for the exercises were not standardised between studies, which could have led to variations in the EMG activity. Several studies examined isometric contractions (Ekstrom et al., 2007), whereas others investigated the whole dynamic movement (Distefano et al., 2009; O'Sullivan et al., 2010). Differences between isometric, concentric and eccentric contractions have been identified with respect to the motor control; for dynamic actions, the recruitment threshold was lower than during isometric contractions (Linnamo et al., 2003). Eccentric exercise is known to produce less EMG activity even though the force output is higher, meaning the mean EMG activity may appear lower when the full movement is analysed (Escamilla et al., 2010). With respect to analysing dynamic movements, some authors chose to just evaluate the concentric phase of the movement (Ayotte et al., 2007; Rabel et al., 2012). When only analysing the concentric phase, the EMG activity represented just half of the rehabilitation exercise. Within the current study, the concentric and eccentric phases were analysed, because the results are required to fully reflect the exercises, which patients will be performing during the rehabilitation process. Furthermore, in a CKC exercise, the co-contraction between the agonists and antagonists is a highly important feature to investigate as it relates to functional movements as well as those seen in sport (Escamilla et al., 2002).

There have also been differences between whether studies have used the overall mean EMG activity (Boudreau et al., 2009), the peak EMG value (Lubahn et al., 2011), or a stated time-frame surrounding the peak (Andersen et al., 2006). The mean EMG activity considers both magnitude and time, taking into account the whole exercise, whereas the peak amplitude represents a single moment (Matheson et al., 2001). Within rehabilitation, high peaks of muscle activation are not necessarily the ideal measure as these peaks may be detrimental for the muscles recovering from an injury so it was deemed more appropriate to examine the mean EMG activity. As seen in the previous section, for this study, the data analysis incorporated the full movement, which related itself to using the mean EMG activity rather than the peak value.

2.8 Muscle Ratios

2.8.1 Vastus Medialis Oblique:Vastus Lateralis Ratio

Patellofemoral pain syndrome is commonly experienced by the younger, physical population (Bizzini et al., 2003). The syndrome presents as anterior knee pain, which can be aggravated by extended periods of sitting and stair climbing (Van Tiggelen et al., 2009). The development of PFPS is multi-factorial (Cowan et al., 2002); however, there appears to be a high focus on the imbalance between the weakened VMO and strong VL, causing the patella to maltrack within the femoral groove (Bolgla et al., 2008). Consequently, within rehabilitation, it is important to strengthen the VMO (Van Tiggelen et al., 2009). It has been suggested the VMO could be targeted through manipulating the active range of motion during quadriceps-based exercises (Banovetz et al., 1996; McConnell, 1986). The VMO acts as a medial patella stabiliser due to its oblique orientation of muscle fibres (Waryasz et al., 2008). This function was hypothesised to become more significant during the final 15^{o} of knee extension due to a 60% increase in force production from the quadriceps (Sakai et al., 2000). There is also less of a lateral restraint from the femoral groove onto the patella towards full knee extension (Toumi et al., 2007).

These may be the theories behind some medical practitioners prescribing mini-squats or terminal knee extensions for PFPS rehabilitation (Lieb & Perry, 1968; Witvrouw et al., 2000). However, there is limited evidence to support this and Tang et al. (2001) actually identified a squat between 0-60^{o} of knee flexion produced the greatest VMO:VL ratio. A different study investigated a squat to 90^{o} and found an overall VMO:VL ratio in excess of 1.2, which showed the VMO to be preferentially activated (Boling et al., 2006). This deep squat exercise would need to be performed with caution during the initial rehabilitation phase though, because it was suggested that patients suffering from PFPS should complete closed kinetic chain exercises between 0-45° of knee flexion due to less patellofemoral stresses during this range (McGinty et al., 2000). It is important that no pain is experienced during exercises because it was identified that pain can inhibit the VMO, and if just 20-30ml of swelling becomes present, the VMO inhibits further, causing the symtoms to heighten (Hopkins et al., 2001).

It was questioned whether the VMO:VL ratio during functional movements was normally close to the value of one for healthy individuals after Worrell et al. (1998) identified large variations in the ratio throughout a cohort of asymptomatic subjects. A number of studies have recognised there is an additional branched nerve supply to the VMO and individual motor points in the vastus medialis and VMO, so neuromuscular activation and co-ordination may not be identical in everyone (Lieb & Perry, 1968; Toumi et al., 2007). Having said this, it is largely assumed within the literature that clinicians need to aim for their athletes to have an equal ratio between the vasti muscles (Sakai et al., 2000; Westfall & Worrell, 1992).

The current research is still controversial for whether concurrent isometric hip adduction during a quadriceps exercise has the ability to increase the VMO:VL ratio (Balogun et al., 2010; Hanten & Schulthies, 1990; Hertel et al., 2004). The varied results could have been due to the differences in the range of knee flexion angles. Earl et al. (2001) utilised a 30° knee flexion angle, whereas Boling et al. (2006) used a larger 90° angle. The method for performing the hip adduction also differed between using a rope and pulley system or squeezing a ball, which could have affected stability at the knee and participants' effort levels (Karst & Jewett, 1993). Some studies used a wall-slide exercise rather than a free squat; this could have altered the difficulty of the task and may have affected the muscular activity due to the position of the lumbar spine (Boling et al., 2006). Hanten & Schulthies (1990) found significant benefits when performing isometric hip adduction, but this was performed in isolation rather than alongside a dynamic squat or other quadriceps exercise.

Wong et al. (2013) investigated a squat with and without hip adduction but their main aim was to compare between surface and fine-wire EMG. It was concluded that the VMO:VL ratio was significantly higher with the inclusion of hip adduction only when measuring with surface EMG. It may be speculated that this occured due to the increased risk of cross-talk between the VMO and adductor magnus. However it was suggested that it is possible to eliminate this if guidelines are adhered to for the positioning of the surface electrodes and inter-electrode distance (Smith et al., 2009).

Instead of using EMG, Baffa et al. (2012) conducted a quantitve magnetic resonance imaging (MRI) study to assess the muscle workload during a squat with isometric hip adduction and abduction. It was concluded that a standard squat to 60^{o} of knee flexion was best to produce significantly greater VMO EMG activity, which was a similar finding to that of Tang et al. (2001). The GMed was found to produce significantly greater EMG activity during a squat with abduction compared to the squat with adduction. No differences in VMO:VL were evident in these two conditions.

The addition of external hip rotation during a quadriceps exercise is another area which has inconclusive results; some studies found no effect on the VMO:VL ratio (Herrington et al., 2006; Karst & Jewett, 1993) whereas a study by Sykes and Wong (2003) concluded this hip position was able to increase the VMO activity. For the latter study, it was suggested that the adductors had a higher electrical activity from being in a stretched position during external rotation, which could have been recorded if the VMO signal became contaminated due to the muscles' close proximity (Herrington et al., 2006). However, as stated before, the varied results suggest it is likely that this cross-talk can be reduced or abolished. Other studies concluded internal tibial and hip rotations were actually more efficient at activating the VMO than external rotation (Cerny, 1995; Lam & Ng, 2001). Varying degrees of rotation and other methodological differences make direct comparisons between studies difficult.

Deficits in hip strength, specifically the hip abductors and external rotators, have also been related to the incidence of PFPS (Cichanowski et al., 2007; Robinson & Nee, 2007), so this is another area which needs to be addressed (de Marche Baldon et al., 2009). Internal rotation of the femur can alter the quadriceps angle (Q angle) and lateral force vector, therefore influencing the position of the patella in the trochlea groove (Powers, 2003). The Q angle is defined as the angle between a line from the ASIS to the centre of the patella, and a line from the centre of the patella to the tibial tuberosity centre (France & Nester, 2001).

One study concluded that pain levels during functional movements decreased to a greater extent when both the quadriceps and hip abductors were strengthened, rather than just the quadriceps (Nakagawa et al., 2008). A recent study actually identified that isolated hip strengthening was capable of decreasing pain and improving function in patients with PFPS (Khayambashi et al., 2012). Therefore, the GMed and GMax EMG activity, as well as the VMO:VL ratio findings from the current study will be beneficial for such patients.

As mentioned before, the VMO:VL ratio is also important when treating patients who are recovering after knee surgery because the VMO was identified as the first quadriceps muscle to undergo atrophy (Sakai et al., 2000). A study which artificially injected 30ml of saline into the knee identified the VMO was still inhibited 210 minutes later. After surgery or injury, the body would not reabsorb the fluid so quickly so this timeframe would be greatly extended (Hopkins et al., 2001).

2.8.2 Hamstrings:Quadriceps Ratio

With regards to the HS:Quads ratio, there have been studies to suggest this is a factor to consider in patients with knee osteoarthritis (Hortobágyi et al., 2005). It was concluded that patients with knee osteoarthritis performed activities of daily life, such as walking and stair climbing, with a larger HS:Quads ratio than a) healthy age-matched controls and b) healthy, younger adults. In terms of the greater EMG activity throughout the osteoarthritis group, it could be speculated this was from a decreased reference value rather than an increased EMG activity during the movements, because pain was experienced at a mean level of 1.83 on a 0-4 point scale. Hortobágyi et al. (2005) suggested focusing on the imbalance between the hamstrings and quadriceps, making the results from the current study important.

The ACL prevents anterior displacement of the tibia on the femur, so is put under pressure when there is an anterior shearing force on the proximal tibia (Herman et al., 2008). Anterior cruciate ligament ruptures account for up to 80% of all knee ligament injuries (Gianotti et al., 2009). The majority of these are the result of a non-contact mechanism of injury, meaning there was no direct blow to the knee at the time of injury (Koga et al., 2010). Females are 4-6 times more likely to suffer from an ACL injury than males competing in the same sports (Hewett et al., 2005). Olsen et al. (2004) found the injury usually occurred when the foot was planted on the ground, combined with knee valgus in a slightly flexed position, and external rotation of the tibia. This is a high-risk position which results in the hip abductors and external rotators being in a mechanically disadvantaged position to protect the knee (Ireland, 1999). This process of events was found to happen within a 40 ms time frame, after initial contact with the ground (Koga et al., 2010). However, if the athlete has correct neuromuscular control, the knee has a greater chance of remaining stable in this type of position, thus preventing injury (Myer et al., 2005). One study identified subjects, who had stronger hip external rotators, to have lower knee anterior shear forces and less external knee adduction (Lawrence et al., 2008). When referring to the anterior shearing force on the tibia, the contraction of the quadriceps produces this force, whereas the hamstrings oppose it by producing a posterior shearing force on the proximal tibia (Stuart et al., 1996). If there is a large imbalance between the two muscle groups and the anterior shear force is large enough, an injury will occur. This highlights the importance of the HS:Quads ratio during exercises; this should be examined as the hamstrings need to be strengthened to produce a counter-movement to the quadriceps.

There has been much speculation as to the causes of the gender differences with ACL ruptures. It was identified that females landed with less knee flexion, more quadriceps activation and lower hamstring contraction levels which could predispose them to injury (Chappell et al., 2007). Berns et al. (1992) focused on the anterior shearing forces in the knee and found the ACL was loaded to a lesser extent as the knee flexion angle increased, therefore, females landing with minimal flexion, exposed the ACL to higher forces. This also relates to the co-contraction of the quadriceps and hamstrings producing the shear forces on the tibia as explained before. Gender differences in coronal plane neuromuscular activity are also evident, with increased knee abduction/adduction moments observed in females predisposed to this injury (Hewett et al., 2005).

The activation of the four quadriceps muscles were also investigated and it was found that females had a decreased medial to lateral quadriceps ratio, potentially increasing the valgus position of the knee (Myer et al., 2005; Palmieri-Smith et al., 2008). The medial muscles provide resistance for abduction at the knee, so a decrease in this activation in females may place unnecessary stress on the ACL (Palmieri-Smith et al., 2009). The larger Q angle may offer a greater angle of pull for the VL; such an increase in the VL activation was seen to increase the anterior shearing forces on the proximal tibia, affecting the load on the ACL (Sell et al., 2007). When considering these theories, it is important to remember that not only do the quadriceps attach into the quadriceps tendon and consequently the patella tendon, some fibres attach directly onto the patella and patella tendon (Figure 1) (Lefebvre et al., 2006; Toumi et al., 2007).

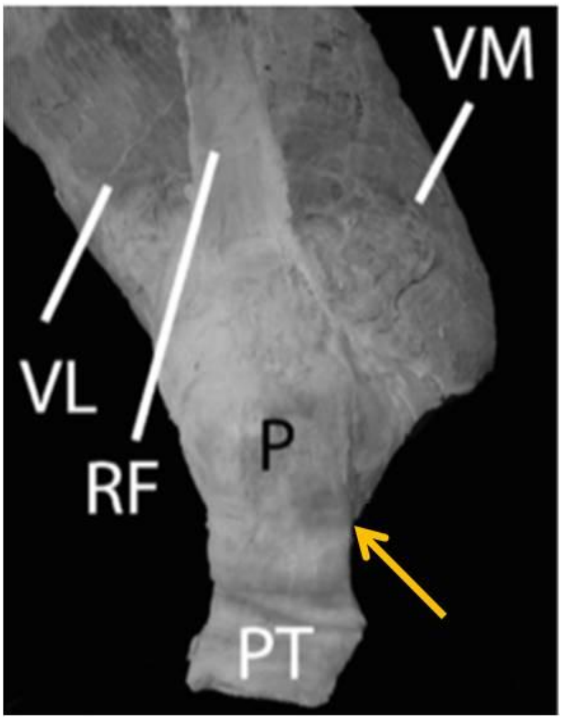

Figure 1. Frontal view during the dissection of the vastus medialis, VL and RF. The arrow depicts where the vastus medialis inserts directly into the patella tendon. Adapted from Toumi et al. (2007, p. 1158).
VM: Vastus Medialis, VL: Vastus Lateralis, RF: Rectus Femoris, P: Patella, PT: Patella Tendon.

In terms of rehabilitation, there is slight controversy between completing OKC or CKC exercises for rehabilitation, even though activities of daily life require both types of movement (Beynnon et al., 1997). Kvist and Gillquist (2001) investigated the anterior tibial translation during both OKC and CKC exercises on injured and healthy subjects; it was concluded there was a larger tibial translation during OKC active knee extensions compared to CKC squatting movements. Similar findings were presented by Yack et al. (1993) and Lysholm and Messner (1995). One thought was that during weight bearing CKC exercises, the tibiofemoral joint had a higher compressive force, thus allowing less tibial translation (Kvist & Gillquist, 2001). In a cadaveric study, this concept was highlighted when a loaded knee produced less anterior-posterior shift of the tibia on the femur, compared to an unloaded knee (More et al., 1993). However, Beynnon et al. (1997) observed no significant differences in tibial translation between a squatting exercise and an OKC extension activity. This difference in results could have been from varied methods of measuring the tibial translation and different protocols for performing the squats. In the study by Kvist and Gillquist (2001), the squat with the centre of mass positioned behind the feet produced the least amount of translation, which was dissimilar to how the squats were performed in the study by Beynnon et al. (1997).

2.8.3 Gluteus Maximus:Biceps Femoris Ratio

During a search of the current literature, no studies were found to specifically investigate the ratio between the GMax and BF during rehabilitation exercises. This is not to say they didn't measure the EMG activity of both muscles, but no relationship in the magnitudes were analysed (Bruno et al., 2008; Chance-Larsen et al., 2010; Kang et al., 2013; Lehman et al., 2004). The current study has chosen to examine the GMax:BF ratio because it was identified that an alteration in the activation of the synergists in a force couple relationship can impair the dominant muscle (Jonkers et al., 2003). In this situation, a weakened GMax can increase the contribution of the hamstrings during hip extension (Chance-Larsen et al., 2010; Lewis et al., 2007), potentially pre-disposing the muscle to injury (Devlin, 2000). Hamstring strains were suggested to be the most prevalent injury amongst footballers and sprinters (Bahr & Mæhlum, 2004), with recurrence rates being as high as 31% (Petersen & Hölmich, 2005).

Previous research has measured the onset timing of the hamstrings, GMax and erector spinae during a prone hip extension task on healthy individuals (Bruno et al., 2008; Chance-Larsen et al., 2010; Lehman et al., 2004). There was no consensus between the findings of these studies as to what constituted as 'normal' muscle onset timing of the three muscles during such a test. However, there was a common trend that the GMax was last to fire. These overall variations could be due to physiological differences between all individuals or because the asymptomatic subjects, classed as healthy participants, actually had underlying motor control issues, which were yet to manifest with injury symptoms (Lehman et al., 2004). A delayed GMax was originally thought to be associated with low back pain (Nadler et al., 2001). It cannot be stated that GMax strengthening will improve the timing issues of the muscle activation (Bruno et al., 2008), but instead, it can affect the patient's gait especially during the loading stance phase, when the GMax should be the principle hip extensor (Lyons et al., 1983). A recent study concluded that adding fluid into the intra-articular hip produced an arthrogenic neuromuscular inhibition of the GMax, thus representing that gluteal strength can further diminish once injured (Freeman et al., 2013). In all such cases where the GMax is weakened so the hamstrings act as the dominant hip extensor, it can be important to preferentially activate and strengthen the GMax. If both muscle groups are strengthened to similar amounts, the imbalance will still remain.

2.9 Gender Differences

From reviewing the literature up until this point, an additional aspect has been identified which requires further investigation; the research is still unclear as to whether gender affects the lower limb muscle activity. Therefore, an additional aim of this thesis was to analyse the effect of gender on the mean EMG activity during the same twenty rehabilitation exercises.

Dywer et al. (2010) observed the total concentric GMax EMG activity during a single-leg squat, lunge and step-up-and-over was greater in females compared to males. For the eccentric GMax activation, females also exhibited greater EMG activity during the single-leg squat and lunge. However, Bouillon et al. (2012) did not identify any significant differences between the genders for the EMG activity of the GMax, GMed, BF and RF during the lunge and step down. In the study by Dywer et al. (2010), greater hip extension range of motion was viewed in the females compared to the males during the exercises; this increased range of motion could have heightened the GMax activity in females (Chumanov et al., 2008). It is important for clinicians to be aware of these findings to ensure rehabilitation incorporates every factor that could affect the strengthening process.

There were no gender differences observed when analysing the EMG activity of the GMed during single-leg exercises in studies by Bouillon et al. (2012), Dwyer et al. (2010) and Zazulak et al. (2005). In opposition to this, other research highlighted that females had lower GMed activity than males, shown by a greater hip adduction torque (Chumanov et al., 2008; Earl et al., 2007). This is therefore an area which needs clarification for future rehabilitation and prehabilitation purposes, especially to prevent hip adduction and knee valgus forces.

Gender differences have also been evident for the RF mean EMG activity during lower limb exercises (Dywer et al., 2010; Zeller et al., 2003) and during the pre-contact period when landing from a jump (Zazulak et al., 2005). This may be due to variations within the pelvic alignment of different subjects, with anterior pelvic tilts being associated with increased hip flexor activity (Tateuchi et al., 2012). Finally, increased BF mean EMG activity in males compared to females was observed when landing from a jump (Chappell et al., 2007; Ebben et al., 2010). This may have implications for the prevention of ACL injuries in females due to the need for the hamstrings to contract to withstand the anterior tibial translation forces produced by the quadriceps (Stuart et al., 1996). The differences in muscle activity need to be investigated for both males and females to ensure exercise rehabilitation is specific for each gender.

Chapter Three: Pilot Testing

3.1 Pilot Study 1: The definition of 'Leg Dominance'

3.1.1 Introduction

There is currently a lack of consensus in the literature between the methods used for determining leg dominance; should it be established on the grounds of strength, personal preference or skill? (Hoffman et al., 1998). Some methods, which have commonly been used to identify leg dominance, include the leg chosen to kick a ball (Ayotte et al., 2007; Lubahn et al., 2011), the leg which produced the greatest distance during a single-leg hop (Nyland et al., 2004), and the limb to recover balance after a perturbation pushing the subject anteriorly (Hoffman et al., 1998). However, questions have arisen when considering the kicking action; is the dominant leg the one used to provide stability and bear the weight? (Hollman et al., 2006), or is it the one which is used to kick the ball? (Hoffman et al., 1998). For the general population and athletes who partake in bilateral sports, such as running, it was reported that the left and right sides showed insignificant differences (McCurdy & Langford, 2005). However, Jacobs et al. (2005) specifically investigated the hip abductor strength and concluded there were differences between sides, on average of approximately 11% in healthy, asymptomatic subjects. Junior footballers were also identified to have inconsistencies in the function and hypertrophy levels between legs, although the study was restricted to the gastrocnemius (Kearns et al., 2001). In order to compare the results of all studies that utilise the concept of leg dominance, these latter two studies suggest it is important to always define the dominant leg in a similar fashion if data is only collected from one limb. This would decrease any immediate discrepancies between studies. Pilot work was therefore completed to determine if subjects consistently recruited the same leg for different tasks.

3.1.2 Method

Subjects attended one testing session and undertook three tests in a randomised order. Three trials were performed for each test. The tests included: kicking a ball, repeatedly hopping on a single leg for five meters and recovering balance after a perturbation pushing the subject anteriorly. Each test was performed three times. Forty three subjects, 25 males and 18 females (mean ± standard deviation, age: 18 ± 0.9 years; height: 169.3 ± 7.3 cm; mass: 66.4 ± 9.5 kg) were included in this pilot work and they conformed to the inclusion criteria for the main study (Section 4.2, page 49). The methodology for each test is listed below.

<u>Kicking Action</u>: A ball was placed on the floor directly in front of the subject, ensuring the positioning did not favour the left or right leg. The subject was then asked to accurately kick the ball straight forwards to the examiner who stood 5 m away. The leg which made contact with the ball was classed as the dominant leg.

<u>Hopping</u>: The hopping task involved the subject standing 5 m away from the examiner. The examiner instructed the subject to repeatedly hop towards them, always taking off and landing on one leg. The same leg had to be used for the full 5 m. This leg was noted as the dominant leg.

<u>Recover Balance</u>: For the perturbation, the subject was pushed forwards from behind with the examiners hand contacting the subject directly between the scapulae. The force applied was just enough to make the subject lean forwards and have to take a step to regain balance. The leg which stepped forwards first was recorded as the dominant side.

<u>Writing Hand</u>: The subjects were also asked which hand they chose to write with.

3.1.3 Data Analysis

The modal result from the three trials for each task was used within the data analysis for each subject. The percentage of subjects who completed the tasks on the right and left leg were calculated for each task.

3.1.4 Results

The analysis identified that 74.4% of subjects preferred to hop on the leg which was used to kick the ball, and 67.4% also regained their balance after a perturbation using the kicking leg. A relatively smaller value of subjects (58.1%) regained their balance on the limb they chose to hop with. The hand the subjects wrote with did not appear to fully represent their dominant leg for the hopping task and to regain balance after a perturbation because 32.6% of subjects used the contralateral leg to the writing hand. However, only 16.3% chose to kick a ball with the opposite leg to the writing hand.

3.1.5 Discussion

The results demonstrated that the greatest percentage of subjects used their kicking leg to also complete the other two tasks so it can be suggested that the kicking leg showed leg dominancy. This opposed the work of Hollman et al. (2006), which stated that the weight bearing side during the kicking action was the dominant leg. The hopping and kicking tasks were easier to repeat and control than the perturbation task. It was difficult to standardise the force used during the perturbation because it was required to be relative to each subject as the body’s inertia had to be overcome in order to produce a movement (McGinnis, 2004). The hopping task differed from other literature as it was based on the preferred leg; other research had recorded the leg which produced the furthest distance travelled (Nyland et al., 2004). The current study performed the test in the previously described way, so that the kicking task incorporated skill, the hopping test involved the leg which the subject perceived as being the dominant side, and the perturbation task showed the leg which stabilised the body after a movement. A limitation with this study was the lack of strength assessment; an assumption was made that the leg chosen to kick a ball and hop with were the stronger side as highlighted in the study by Hoffman et al. (1998).

3.1.6 Conclusion

As the results have shown, the greatest percentage of subjects also used their kicking leg for the hopping task and to provide balance. Therefore, within the rest of this thesis, the leg used to kick a ball was adopted as the definition for limb dominance.

3.2 Pilot Study 2: Reliability of the EMG results

3.2.1 Introduction

When considering the reliability of measurements, there are various options for which statistical tests to perform. Fauth et al. (2010) suggested using ICC and inter coefficients of variance for within session trial to trial reliability; low values are ideal to show consistency. Studies have found high reliability for isometric quadriceps contractions between sessions, with an ICC of 0.99 (Fauth et al., 2010; McCarthy et al., 2008). It was also suggested that the reliability was not affected by a familiarisation visit so these were not seen to be crucial (McCarthy et al., 2008). Specifically to the hamstrings, it was identified that the BF produced more reliable EMG activity than the semitendinosis (Kellis & Katis, 2008).

This pilot study is split into two parts:

- Section 1) The reliability of the MVICs between days
- Section 2) The reliability of the within-subject EMG activity during the twenty rehabilitation exercises.

Section 1: The Reliability of the MVICs Between Days

3.2.2 Method

Seven subjects were used for this section of the pilot study (females, n = 3, males, n = 4; mean ± standard deviation, age: 18.5 ± 0.6 years; height: 169.3 ± 4.3 cm; mass: 68.4 ± 9.1 kg). The skin was cleaned with an alcohol wipe prior to the application of the electrodes. An eight-channel surface electromyographic system (Biometrics UK Ltd DataLOG) was used to collect data from the muscles on the dominant leg (GMax, GMed, BF, RF, VMO and VL). Leg dominancy was determined by the leg used to kick a football as previously discussed. SENIAM guidelines were adhered to for the positioning of the electrodes on the muscles. Pre-amplified SX230 disc electrodes (Ag/AgCl), with an inter-electrode distance of 20 mm, were used. The electrode diameter measured 10 mm. The ground reference electrode (R206) was applied to the ipsilateral pisiform. The EMG signal was low and high pass filtered between 20-450 Hz.

The subjects performed three trials of each MVIC. Each test lasted 5 seconds, during which standardised verbal encouragement was given. There were two types of MVIC tests for each muscle, with varying subject positions so it could be determined which method was 1) more reliable and 2) produced the greater EMG activity. The subject positioning and location of the resistance for the two tests per muscle are presented in Table 2. The resistance was applied using a non-elasticated belt which was attached to the bed. One week later, the subjects returned for a second testing session; the same procedure was repeated.

Table 2. Descriptions of the two types of MVIC tests for each muscle.

Muscle	Test	Position of resistance belt
GMax	a) Lying prone, knee flexed to 90°, isometric hip extension	a) Distal thigh, just proximal to the popliteal crease
	b) Standing, knee flexed to 90°, isometric hip extension	b) Distal thigh, just proximal to the popliteal crease
GMed	a) Side-lying, in 20° of abduction, isometric hip abduction	a) One inch proximal to the lateral femoral condyle
	b) Standing, in 20° of abduction, isometric hip abduction, measured non weight bearing limb	b) One inch proximal to the lateral femoral condyle
BF	a) Lying prone, knee flexed to 45°, isometric knee flexion	a) One inch proximal to the lateral malleolus
	b) Lying prone, knee flexed to 45°, isometric knee flexion plus hip extension	b) One inch proximal to the lateral malleolus and distal thigh
Quadriceps: RF, VMO and VL	a) Seated, isometric knee extension at 45°	a) One inch proximal to the lateral malleolus
	b) Seated, isometric knee extension at 60°	b) One inch proximal to the lateral malleolus

MVIC: Maximal Voluntary Isometric Contraction, GMax:Gluteus Maximus, GMed: Gluteus Medius, BF: Biceps Femoris, RF: Rectus Femoris, VMO: Vastus Medialis Oblique, VL: Vastus Lateralis.

3.2.3 Data Analysis

The data was RMS with time-intervals of 2 ms. The 100 ms of EMG activity surrounding the peak value was averaged and used throughout the data analysis. This included data from 50 ms prior to the peak EMG activity, and 50 ms after. Paired sample t-tests were used to examine any differences between the results from the first and second testing sessions. Subsequently, ICCs and SEM were calculated to estimate the reliability. Paired sample t-tests were also used to compare the EMG activity between tests (a) and (b) for each of the muscles to determine which type of test produced the greatest muscle activation. Significance was determined with alpha levels set to $p < 0.05$.

3.2.4 Results

The standing GMax test re-test was the only MVIC which produced significantly different results at the two sessions ($p = 0.008$). The ICC was lowest for the standing GMax test (ICC = 0.40) and highest during the RF test in 60° of knee flexion (ICC = 0.99) (Table 3). The SEM values differed between 0.79% and 7.85% MVIC (Table 3).

Table 3. Reliability of each of the MVIC test re-tests.

Test	ICC	SEM (%)
(a) GMax - lying prone	0.95	1.53
(b) GMax - standing	0.60	5.66
(a) GMed - side-lying	0.87	1.26
(b) GMed - standing	0.40	5.31
(a) BF at 45°	0.96	0.71
(b) BF with hip extension	0.74	2.23
(a) RF at 45°	0.82	6.32
(b) RF at 60°	0.99	0.79
(a) VMO at 45°	0.78	7.85
(b) VMO at 60°	0.89	3.98
(a) VL at 45°	0.80	5.36
(b) VL at 60°	0.96	1.52

MVIC: Maximal Voluntary Isometric Contraction, ICC: Intraclass Correlation Coefficients, SEM: Standard Error of Measurement GMax:Gluteus Maximus, GMed: Gluteus Medius, BF: Biceps Femoris, RF: Rectus Femoris, VMO: Vastus Medialis Oblique, VL: Vastus Lateralis.

The paired sample t-tests did not identify any significant differences between test (a) and test (b) for the amount of EMG activation for each muscle ($p > 0.05$) (Table 4).

Table 4. Results comparing the EMG values between the two types of MVIC tests for each muscle.

Test a	Test b	Significance level between tests (p value)
GMax lying prone	GMax standing	0.36
GMed side-lying	GMed standing	0.29
BF at 45°	BF with hip extension	0.27
RF at 45°	RF at 60°	0.72
VMO at 45°	VMO at 60°	0.07
VL at 45°	VL at 60°	0.40

MVIC: Maximal Voluntary Isometric Contraction, GMax: Gluteus Maximus, GMed: Gluteus Medius, BF: Biceps Femoris, RF: Rectus Femoris, VMO: Vastus Medialis Oblique, VL: Vastus Lateralis.

3.2.5 Discussion

The present findings identified that each of the MVIC tests had varied reliability, ranging from poor through to very high reliability, according to the scale used by Mathur et al. (2005). It was encouraging that for each muscle, at least one of the two tests produced a minimum of a ‘high reliability’ result (ICC > 0.87). For both the

GMax and GMed, the lying tests appeared more reliable than the standing tests which could have been due to the stability of the pelvis (Jung et al., 2012).

For the BF tests, the isolated knee flexion produced a higher ICC and lower SEM value than the knee flexion combined with hip extension, which represented the first method was more reliable (Kellis & Katis, 2008). This may have been due to it being more difficult to standardise the double contraction (test b) as it was more complex for the subjects to perform. Finally, the quadriceps contraction in 60° of knee flexion produced more reliable results than the test in 45° knee flexion for the RF, VMO and VL. As there were no differences in the EMG activity between the two angles, the more reliable 60° test was used throughout the rest of this thesis for the MVIC of the quadriceps, similarly to the work by Farrokhi et al. (2008).

3.2.6 Conclusion

Due to the increased reliability, within the main testing for this thesis, the lying MVIC tests were used for the GMax and GMed, the MVIC incorporating knee flexion at 45° was used for the BF and finally, the MVIC at 60° of knee flexion was used for the RF, VMO and VL.

Section 2: Within-subject Reliability During the Rehabilitation Exercises

3.2.7 Method

Ten subjects were used for the second part of this pilot study (females, n = 5, males, n = 5; mean ± standard deviation, age: 18.7 ± 0.4 years; height: 170.1 ± 5.0 cm; mass: 69.6 ± 7.3 kg). An eight-channel surface electromyographic system (Biometrics UK Ltd DataLOG) was used to collect data from the GMax, GMed, BF, RF, VMO and VL on the dominant leg. The skin was cleaned with an alcohol wipe prior to the application of the electrodes. Electrodes were positioned following the SENIAM guidelines. Pre-amplified SX230 disc electrodes (Ag/AgCl), with an electrode diameter of 10 mm, were used. The inter-electrode distance measured 20 mm. The ground reference electrode (R206) was applied to the ipsilateral pisiform. The EMG signal was low and high pass filtered between 20-450 Hz.

The subjects performed three trials of each exercise. As there were twenty exercises, these were split over three sessions in a randomised order to avoid bias results from fatigue. The twenty exercises were categorised into different stages of rehabilitation (Table 5) depending upon the type of movement and degree of load bearing on the dominant leg, as based on the criteria in Table 1 (Section 2.1.1, page 14). Photographs of all exercises can be viewed in Appendix 1.

Table 5. The exercises categorised into their corresponding stage of rehabilitation.

Early Stage	**Intermediate Stage**	**Late Stage**	**Performance Enhancing**
- Quadriceps Setting - Straight Leg Raise - Side-lying Hip Abduction - Standing Hip Abduction - Prone Hip Extension	- Mini Squat - Full Squat - Lunge - Bridge - Single-leg Bridge - Raised Bridge	- Step Up - Step Down - Lateral Step Up - Hip Hitch - Single-leg Squat	- Counter Movement Jump - Single-leg Vertical Jump - Weighted Squat - Deadlift

3.2.8 Data Analysis

The EMG data was RMS with 2 ms time intervals. The mean EMG activity during the full 2 second exercise was used during the analysis, thus incorporating both the concentric and eccentric phases. The ICCs were calculated across the three trials for within-subject reliability.

3.2.9 Results

The ICCs varied between 0.805 and 0.998 showing high within-subject reliability. Table 6 provides the results for each muscle and each exercise.

Table 6. Within-subject reliability for each of the exercises.

Exercise	ICC for each muscle					
	GMax	GMed	BF	RF	VMO	VL
Quadriceps Setting				0.916	0.932	0.932
Straight Leg Raise				0.987	0.989	0.989
Side-lying Hip Abduction		0.993				
Standing Hip Abduction		0.805				
Prone Hip Extension	0.987		0.826			
Mini Squat	0.986	0.927	0.989	0.957	0.992	0.979
Full Squat	0.964	0.880	0.998	0.904	0.993	0.987
Lunge	0.984	0.970	0.982	0.929	0.992	0.984
Bridge	0.986	0.960	0.996	0.977	0.792	0.925
Single-leg Bridge	0.989	0.975	0.932	0.891	0.986	0.989
Raised Bridge	0.882	0.850	0.938	0.930	0.929	0.881
Step Up	0.950	0.975	0.986	0.989	0.984	0.990
Step Down	0.973	0.947	0.974	0.971	0.993	0.988
Lateral Step Up	0.978	0.960	0.964	0.953	0.974	0.972
Hip Hitch		0.967				
Single-leg Squat	0.985	0.963	0.981	0.969	0.987	0.967
Counter Movement Jump	0.983	0.985	0.975	0.990	0.995	0.981
Single-leg Vertical Jump	0.992	0.875	0.951	0.964	0.970	0.983
Weighted Squat	0.991	0.964	0.989	0.997	0.994	0.986
Deadlift	0.986	0.970	0.992	0.991	0.985	0.970

ICC: Intraclass Correlation Coefficient, GMax: Gluteus Maximus, GMed: Gluteus Medius, BF: Biceps Femoris, RF: Rectus Femoris, VMO: Vastus Medialis Oblique, VL: Vastus Lateralis.

3.2.10 Conclusion

All of the exercises produced a high ICC showing adequate reliability throughout all of the muscles. These exercises can therefore be used within the main study of this thesis in the knowledge that the within-subject repeatability is sufficient.

Chapter Four: Methodology

4.1 Study Design

This study used a one-way repeated measures design. The twenty rehabilitation exercises were the independent variables whilst the EMG activity acted as the dependant variable.

4.2 Participants

Eighteen physically active volunteers participated in the study. The group consisted of nine females and nine males (mean ± standard deviation, age: 20 ± 1.3 years;

height: 168.1 ± 9.7 cm; mass: 64.1 ± 9.8 kg). Participants were required to be between 18-25 years old, and participate in over an hour of exercise each week. Participants were excluded if they had a history of surgery to the lower limb, a knee injury in the past 12 months, history of a musculoskeletal injury to the lower limb in the previous 6 months, suffered from any central or peripheral neurological conditions, were allergic to adhesive tape, or were pregnant. A brief medical history and physical examination was performed by a graduate Sports Therapist. Ethical approval was obtained from the University Ethics Committee and written informed consent was provided by each participant.

4.3 Instrumentation

An eight-channel surface electromyographic system (Biometrics UK Ltd DataLOG W4X8) was used to collect data from the GMax, GMed, BF, RF, VMO and VL on the dominant leg. Pre-amplified SX230 disc electrodes (Ag/AgCl), with an inter-electrode distance of 20 mm, were used. The electrode diameter measured 10 mm. The ground reference electrode (R206) was applied to the ipsilateral pisiform. The EMG signal was low and high pass filtered between 20-450 Hz. A twin axis electrogoniometer (SG110) was used to measure the knee flexion angle during all exercises.

4.4 Procedure

All data was collected in a Human Physiology laboratory. Subjects wore shorts, t-shirt and were barefoot to prevent any influence from different footwear. The participants' dominant leg was determined by them accurately kicking a ball 5 metres; the leg which performed the kicking motion was classified as the dominant side (Balogun et al., 2010) (right = 16, left = 2). This method was chosen as a result of pilot work (Section 3.1, page 41). The 20 exercises (Table 5) were randomly split over three testing sessions. Seven exercises were, therefore, completed at each session, with one visit having a 'no exercise' alternative so there were 21 options in total. Exercises were randomised so to avoid order bias arising due to fatigue. At the beginning of each session, the examiner demonstrated the exercises which would be completed that day. Subjects practiced each exercise three times to limit the learning effect once data collection was underway (Janwantanakul & Gaogasigam, 2005). During the practice trials, verbal and tactile feedback was given to ensure correct technique was displayed. With regards to the verbal feedback, the examiner commented on the subject maintaining a flat back, or keeping hips level, for example. After permission was sought from the subject for tactile feedback to be given by the examiner, the subjects were moved into the correct position for the desired technique by the examiner making physical contact with the limb or tilting the subject's head up, for example. This method of feedback stimulated the mechanoreceptors in an attempt to provide a cue for increased movement coordination.

The skin was shaved and cleaned with an alcohol wipe prior to the application of the electrodes. The SENIAM guidelines were followed for the electrode positioning on the following muscles. The GMax electrode was located 50% of the way between the sacrum and greater trochanter in the direction of the muscle fibres. The electrode for the GMed was placed half way along the line between the iliac crest and greater trochanter. For the BF, the electrode was positioned at 50% on the line between the lateral tibial epicondyle and the ischial tuberosity. The RF electrode was placed half way between the superior patella and the anterior superior iliac spine (ASIS). Electrode placement for the VMO was 20% of the distance between the medial patella and ASIS, orientated perpendicular to the line between these landmarks. Finally, the VL electrode was a third of the distance between the lateral patella and ASIS, orientated along the perceived line.

The Biometrics UK twin axial electrogoniometer was placed on the dominant leg to monitor sagittal plane knee kinematics. The proximal arm was placed along the line from the lateral epicondyle of the femur to the greater trochanter, whilst the distal arm was on the line between the head of the fibula and the lateral malleolus. Subjects performed two mini squats so the examiner could ensure the equipment was functioning correctly by viewing the trace from the electrogoniometer on the computer.

In order to normalise the EMG data, MVICs were performed for each muscle (Table 7). Photographs can be viewed in Appendix 2. Each muscle was tested three times, for 5 seconds each. There was a 30 second rest between trials and a minute rest between exercises. Verbal encouragement was provided during the MVICs. The same examiner completed all of the testing in order to standardise the wording and volume of the encouragement.

Table 7. MVIC test position for each muscle.

Muscle	Test	Position of resistance belt
GMax	Lying prone, knee flexed to 90°, isometric hip extension	Distal thigh, just proximal to the popliteal crease
GMed	Side-lying, in 20° of abduction, isometric hip abduction	One inch proximal to the lateral femoral condyle
BF	Lying prone, knee flexed to 45°, isometric knee flexion	One inch proximal to the lateral malleolus
RF, VMO and VL	Seated, isometric knee extension at 60°	One inch proximal to the lateral malleolus

MVIC: Maximal Voluntary Isometric Contraction, GMax: Gluteus Maximus, GMed: Gluteus Medius, BF: Biceps Femoris, RF: Rectus Femoris, VMO: Vastus Medialis Oblique, VL: Vastus Lateralis.

4.5 Exercises

Subjects were given a 10 second rest period between the trials of each exercise and a 2 minute rest period between the different exercises in order to prevent fatigue (Willardson, 2006). Three trials were completed for each exercise and no encouragement was given during the exercises. A waist belt was worn which contained the DataLOG box and wires were tucked away to avoid interference and minimise cable movement when subjects were performing the exercises. The DataLOG machine acted via Bluetooth so there were no wires connected between the subject and computer.

4.5.1 Quadriceps Setting

Subjects were supine with their hips and knees extended. A foam cylinder with a 12 cm diameter was placed under the knees. Subjects were instructed to straighten their dominant leg and isometrically contract for 5 seconds. No external resistance was applied. The 5 seconds were counted down by the examiner but no verbal encouragement was given.

4.5.2 Straight Leg Raise

Subjects lay supine with both knees extended. Using trigonometry, it was determined that in order for the hip to abduct by 30°, the foot had to be raised to a height equal to 50% of the subject's leg length. For the purposes of this, leg length was defined as the distance between the ASIS and lateral malleolus. The height of the foot was controlled by having an adjustable hurdle, which the subject had to lightly touch with their anterior joint line of the ankle. The ankle was kept in a neutral position and knee fully extended throughout. The leg was lowered back down in a controlled manner.

4.5.3 Side-lying Hip Abduction

Subjects lay on their non-dominant side, flexing this hip and knee to 90° to stabilise the pelvis. The dominant knee was extended and in 0° of hip extension. This limb was raised to 30° of hip abduction, again using the trigonometry calculation for the height of the hurdle. Subjects were instructed to lightly touch the hurdle with their lateral malleolus, ensuring their hips remained stacked and in a neutral position. The dominant leg was then returned to the initial starting position.

4.5.4 Standing Hip Abduction

Subjects stood on their non-dominant limb so the dominant leg was in a non weight bearing position. The dominant hip was abducted until the lateral malleolus lightly touched an upright wooden bar at 30° of abduction, and then subjects returned the leg to the starting position. The same '50% of leg length' calculation was used to determine the distance between the neutral starting position for the dominant foot and the position of the wooden bar.

4.5.5 Prone Hip Extension

Subjects lay prone with both knees extended. The hurdle was once again used so the hip extension was performed to a 30° angle. The knee remained in a fully extended position throughout the exercise; at the highest point, the calcaneus lightly touched the hurdle. Then subjects lowered their leg back down to the starting position.

4.5.6 Mini Squat

Subjects stood with feet shoulder-width apart. Hip, knees and ankles were flexed in a squatting motion until the subjects reached 45° of knee flexion. Subjects then returned to the upright position. The examiner instructed them to keep their chest up, weight over the heels and not to let the knees drop into a valgus position.

4.5.7 Full Squat

The subjects adopted the same starting position and technique as for the Mini Squat; however, this movement was completed to 90° of knee flexion.

4.5.8 Lunge

Subjects stood with their feet shoulder width apart. They stepped forwards with their dominant leg; this was classed as the starting position. The distance between the feet equated to the length of their leg, measured as the distance between the ASIS and lateral malleolus. This method was evaluated during pilot testing and was found to produce a 90° angle on the dominant knee when at the lowest lunge position. One repetition consisted of being in the starting position with feet apart and then flexing knees until the non-dominant knee lightly touched the floor. Subjects raised back up to the starting position ensuring to keep the trunk upright throughout.

4.5.9 Bridge

Subjects lay supine with both knees bent to 90° and feet flat on the floor. Hips were raised off the floor until a straight line was made between their shoulders and knees. Subjects then lowered their hips back to the starting position.

4.5.10 Single-leg Bridge

Subjects were supine with their dominant knee bent to 90° and foot flat on the floor. The non-dominant leg was extended and raised off the floor by 2 inches. The hips were raised up using only the dominant limb and then lowered.

4.5.11 Raised Bridge

Subjects lay supine with both feet flat on a 30 cm bench. Both knees were flexed to 90°. The bridge exercise was completed as previously described.

4.5.12 Step Up

Subjects placed their dominant foot on a 30 cm bench, whilst keeping their non-dominant leg in an extended position with slight dorsiflexion at the ankle. The emphasis was on using the muscles of the dominant limb to extend the hip and knee in order to raise the body up. The subjects then lowered back down to the starting position. The non-dominant limb was not placed on the bench during the exercise.

4.5.13 Step Down

Subjects stood on a 30 cm bench with all their weight on the dominant limb. The non-dominant hip was slightly flexed so the leg hung in front of the bench. The body was lowered down by flexing the dominant knee until the non-dominant heel lightly touched the floor then the subject returned to the upright position.

4.5.14 Lateral Step Up

Subjects stood side on to a 30 cm bench. Their feet were aligned shoulder width apart, with the dominant limb being placed on the bench. Using the dominant leg, subjects raised up extending their knees and hips then lowered back down. The non-dominant limb was not in contact with the bench at any point during the exercise.

4.5.15 Hip Hitch

Subjects stood sideways on the bench with their dominant leg. The non-dominant limb was overhanging the side of the bench. Subjects lowered this side of the pelvis so the non-dominant leg dropped below the level of the bench so to adduct the dominant hip. The non-dominant hip was then hitched up to return to the starting position. The knees were maintained in an extended position throughout the exercise. This exercise incorporated as much pelvic movement as the subjects were able to perform.

4.5.16 Single-leg Squat

Subjects stood on their dominant leg with their non-dominant hip flexed to 45°. The dominant hip and knee was flexed in a squatting motion until the knee reached a 60° angle. The hip, knee and ankle on the dominant side were extended to return the subject to a standing position.

4.5.17 Counter Movement Jump

Subjects stood with feet shoulder width apart. The examiner instructed subjects to aim for maximal height during the flight phase. The take-off and landing were both two-footed. Knee flexion angles were analysed using data from the electrogoniometer. The depth of knee flexion during the counter movement phase was not controlled due to pilot testing showing that when having to squat to a specific depth prior to take-off, the exercise became disjointed and did not reflect a smooth, functional movement.

4.5.18 Single-leg Vertical Jump

Subjects stood on their dominant limb and again, were instructed to aim for maximal height during the flight phase. This exercise was performed by taking off and landing on the dominant leg. Subjects did not travel during the jump as all emphasis was on the vertical component. Knee flexion angles were analysed using data from the electrogoniometer for the same reasons as the counter movement jump.

4.5.19 Weighted Squat

Subjects stood in an upright position. A squat rack was used for safety purposes with the 20 kg Olympic barbell, which was loaded with Olympic disk weights so the total external resistance equalled 40% of the subjects' body mass. The barbell rested on the trapezius muscle as seen in the conventional back-squat technique. In a controlled manner, subjects flexed their knees and hips, lowering down into a squatted position until the knee angle reached 90°. Subjects returned to the starting, upright position. The examiner instructed the subjects to keep their back flat and not to let their knees deviate into a valgus position.

4.5.20 Deadlift

A 20 kg Olympic barbell and Olympic disks were used during this exercise. Again, the total weight of the barbell equalled 40% of the subjects' body mass. Subjects stood with feet shoulder width apart and used a prone grip with hands just lateral to the legs. The conventional bent leg deadlift technique was adopted; knees and hips began in a flexed position with weight distributed posteriorly. Knees and hips were fully extended in the upright position, lifting the bar to mid-thigh level with elbows extended. Subjects then returned to the starting position, ensuring the back remained flat.

4.5.21 Standardised Protocols Adopted For All Exercises

Arms were placed across the chest for the supine exercises to ensure the movements were isolated to the lower limb. Hands were placed on the iliac crests during the standing exercises, except those involving the barbell. For all the squat-based exercises, the knee flexion angle was measured against the height of a hurdle during the practice attempts so subsequent trials were standardised by the subjects squatting until their buttocks lightly touched the bar. A metronome set to 60 bpm was used. For all exercises except the quadriceps setting, the concentric and eccentric phases lasted 1 second each. For example, subjects reached the lowest squatting position after 1 beat and had returned to the upright position by the next beat. With regards to the counter movement jump and single-leg vertical jump, subjects took 1 second to reach the lowest point during the counter movement section then landed by the next beat. If subjects lost balance or were unable to correctly complete the exercise, the data was discarded and the trial was repeated.

4.6 Data Analysis

4.6.1 Normalised EMG Activity

All data were rectified then smoothed using the root-mean-square statistical measure. Two millisecond moving time frames were used for this algorithm. For the MVICs, 100 ms around the peak amplitudes (50 ms prior to the peak value and 50 ms after the peak value) were averaged for use within the data analysis (Figure 2). For each dynamic exercise trial, the mean was calculated for 2 seconds after the onset of muscle activity; thus incorporating both the concentric and eccentric phases (Figure 3). The quadriceps setting exercise differed and used the mean over the full 5 second isometric contraction.

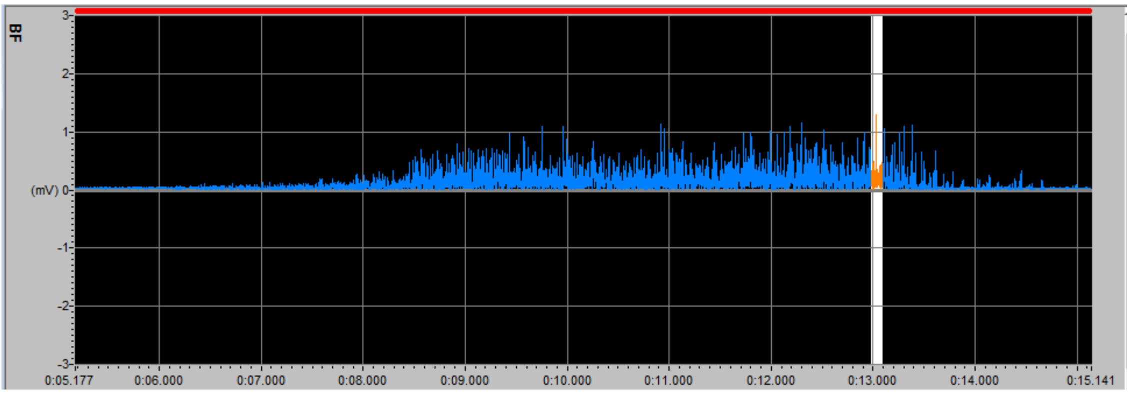

Figure 2. A sample EMG trace for the BF activity during the hamstring MVIC. The highlighted section indicated the 100 ms surrounding the peak value, which was used within the data analysis.

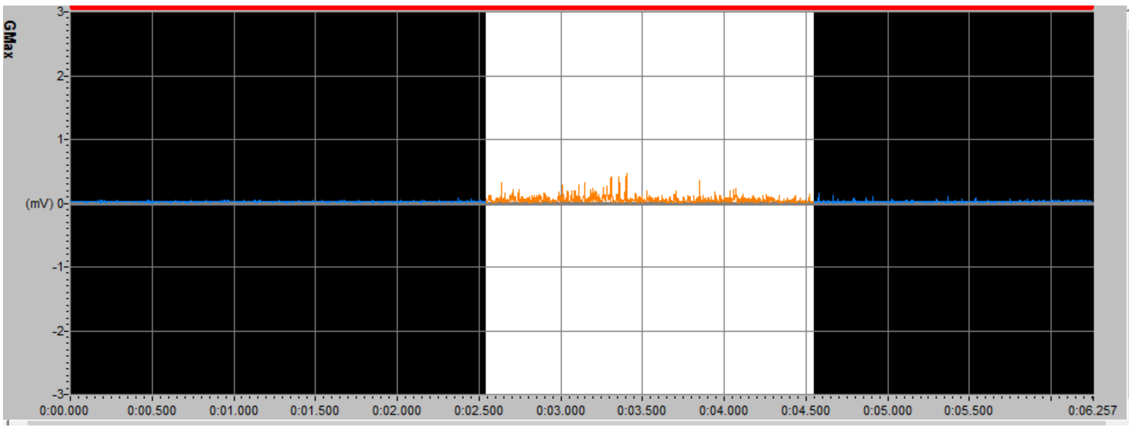

Figure 3. A sample EMG trace for the BF activity during a single-leg bridge. The highlighted section indicates the two second period during which the data was analysed.

The EMG values were normalised using the following equation:

$$\text{Normalised EMG (\% MVIC)} = \frac{\text{EMG value for exercise}}{\text{EMG value for MVIC}} \times 100$$

where the 'EMG value for exercise' was the mean amplitude throughout the 2 second exercise and the 'EMG value for MVIC' was the mean amplitude throughout the 100 ms surrounding the peak magnitude (50 ms either side of the peak).

The normalised EMG values for all of the subjects were averaged for each exercise; these were the mean values used to rank the exercises on the continuum.

Using Statistical Package for the Social Sciences Version 20 (SPSS Inc., Chicago, IL), a one-way repeated measures analysis of variance (ANOVA) was performed using general linear models, to determine any significant difference in muscle activation between the twenty exercises. Post hoc tests for pair-wise comparisons were conducted to determine the points of statistical significance. Bonferroni adjustments were performed to reduce the probability of type I errors. There were 190 possible pair-wise comparisons so alpha levels were set to $p < 0.05 / 190$. However, throughout the thesis, SPSS Bonferroni adjusted p-vales were quoted, thus for significance, $p < 0.05$. The majority of the data were found to be normally distributed using the Shapiro-Wilk test. Other data had a positively skewed frequency distribution so the data underwent logarithmic transformation. The Shapiro-Wilk test was performed after the transformation to confirm all data were normally distributed.

4.6.2 Muscle Ratios

The ratios were calculated for each exercise for every subject then averaged for the whole sample size. The calculations followed a similar method used by Begalle et al. (2012) and Ebben et al. (2009).

The VMO:VL ratio was calculated using the following equation:

$$\text{VMO:VL ratio} = \frac{\text{VMO \% MVIC}}{\text{VL \% MVIC}}$$

where the 'VMO % MVIC' was the normalised EMG activity for the VMO, averaged across the three trials for each subject and the 'VL % MVIC' was the normalised EMG activity for the VL, again, averaged across the three trials for each subject.

The HS:Quads ratio was calculated using the following equation:

$$\text{HS:Quads ratio} = \frac{\text{BF \% MVIC}}{(\text{RF \% MVIC} + \text{VMO \% MVIC} + \text{VL \% MVIC}) / 3}$$

where the 'BF % MVIC' was the normalised EMG activity for the BF, averaged across the three trials for each subject and the 'RF % MVIC', 'VMO % MVIC' and 'VL % MVIC' were the normalised EMG activity for the respective muscles, again, averaged across the three trials for each subject. From this equation, it can be seen that the RF, VMO and VL % MVIC were averaged to represent the activity of the quadriceps, similarly to in the study by Begalle et al. (2012).

The GMax:BF ratio was calculated using the following equation:

$$\text{GMax:BF} = \frac{\text{GMax \% MVIC}}{\text{BF \% MVIC}}$$

where the 'GMax % MVIC' was the normalised EMG activity for the GMax, averaged across the three trials for each subject and the 'BF % MVIC' was the normalised EMG activity for the BF, again, averaged across the three trials for each subject.

4.6.3 Gender Differences

The normalised mean EMG data for all of the exercises were categorised into two groups dependent upon the gender of the subject. Independent samples t-tests were performed on this data comparing the mean EMG activity for the males versus the females. All twenty exercises were analysed. Significance levels were set to $p < 0.05$.

4.6.4 Knee Flexion Angles

The maximum knee flexion angle prior to take-off was recorded for each subject for the counter movement jump and single-leg vertical jump trials. Descriptive statistics were calculated to measure the mean and standard error. A paired samples t-test was conducted to determine the effects of the type of jump (counter movement jump vs. single-leg vertical jump) on knee flexion angle prior to take off. The mean and standard error values were also calculated separately for males and females during the two jumps, step up, step down and lateral step up. Independent samples t-tests were used to compare the male and female knee flexion angles for each exercise, in order to determine if the knee flexion angles were dependent upon gender. Alpha levels were again set to $p < 0.05$.

Chapter Five: Results

Due to the nature of this study analysing six muscles across twenty exercises, the tables and figures in the results section contain high amounts of data. Unfortunately, it was therefore inevitable that some tables span across a number of pages. Where possible, this has been avoided.

5.1 Normalised EMG Activity

The mean EMG activity for each muscle, during the exercises, was expressed as a percentage of the MVIC, as displayed in Table 8 and Table 9. For each muscle, there were significant differences in the EMG activity between the twenty exercises so Bonferroni post-hoc tests were performed to identify where the significance existed.

Table 8. Normalised mean EMG activity in ranked order for the GMax, GMed and BF.

Ranking of muscle activity (highest to lowest)	GMax		GMed		BF	
	Exercise	EMG activity (% MVIC ± SE)	Exercise	EMG activity (% MVIC ± SE)	Exercise	EMG activity (% MVIC ± SE)
1	Single-leg Vertical Jump[a]	45.6 ± 4.9	Single-leg Vertical Jump[j]	51.5 ± 4.0	Counter Movement Jump[r]	59.2 ± 10.2
2	Counter Movement Jump[b]	36.1 ± 5.1	Single-leg Squat[k]	35.9 ± 4.0	Single-leg Vertical Jump[s]	50.0 ± 7.0
3	Single-leg Bridge[c]	33.6 ± 3.9	Single-leg Bridge[k]	35.0 ± 3.9	Single-leg Bridge[t]	42.3 ± 3.3
4	Lateral Step Up[d]	29.8 ± 3.7	Hip Hitch[m]	34.7 ± 6.0	Deadlift[u]	40.8 ± 8.1
5	Step Down[d]	29.1 ± 4.3	Step Down[k]	32.1 ± 3.5	Raised Bridge[v]	36.0 ± 2.6
6	Step Up[d]	27.3 ± 3.1	Counter Movement Jump[k]	31.7 ± 2.8	Weighted Squat[w]	35.6 ± 7.6
7	Single-leg Squat[e]	26.2 ± 2.9	Lateral Step Up[k]	30.5 ± 3.2	Single-leg Squat[w]	33.4 ± 5.6
8	Lunge[f]	20.0 ± 3.1	Step Up[k]	27.3 ± 2.1	Lunge[w]	32.2 ± 6.1
9	Prone Hip Extension[g]	19.9 ± 2.9	Side-lying Hip Abduction[n]	26.1 ± 2.1	Bridge	30.5 ± 4.0
10	Deadlift[h]	18.5 ± 1.8	Bridge[p]	16.8 ± 3.1	Prone Hip Extension	29.7 ± 3.7
11	Bridge[g]	16.3 ± 2.6	Lunge[q]	16.8 ± 2.2	Step Down[w]	25.3 ± 3.1
12	Hip Hitch[g]	16.2 ± 2.8	Deadlift[p]	15.8 ± 2.2	Lateral Step Up	23.7 ± 2.8
13	Raised Bridge[g]	12.4 ± 2.1	Weighted Squat[p]	11.0 ± 1.2	Full Squat[w]	21.8 ± 2.5
14	Weighted Squat[g]	11.7 ± 1.9	Raised Bridge	9.5 ± 1.7	Step Up	21.4 ± 2.6
15	Full Squat[g]	9.4 ± 1.1	Standing Hip Abduction	9.2 ± 2.0	Mini Squat	14.2 ± 3.4
16	Mini Squat	5.1 ± 0.7	Full Squat[p]	8.6 ± 1.3		
17			Mini Squat	5.3 ± 0.7		

GMax: Gluteus Maximus, GMed: Gluteus Medius, BF: Biceps Femoris, MVIC: Maximal Voluntary Isometric Contraction, SE: Standard Error.
a. The single-leg vertical jump produced significantly greater GMax EMG activity than all the other exercises ($p < 0.05$) excluding the single-leg bridge and counter movement jump.
b. The counter movement jump produced significantly greater GMax EMG activity than the lunge ($p < 0.001$) and deadlift through to the mini squat ($p < 0.05$).
c. The single-leg bridge produced significantly greater GMax EMG activity than the deadlift through to the mini squat ($p < 0.05$).
d. The lateral step up, step down and step up produced significantly greater GMax EMG activity than the hip hitch through to the mini squat ($p < 0.05$).
e. The single-leg squat produced significantly greater GMax EMG activity than the bridge ($p = 0.030$) and raised bridge through to the mini squat ($p < 0.05$).
f. The lunge produced significantly greater GMax EMG activity than the full squat ($p = 0.024$) and mini squat ($p < 0.001$).
g. These exercises produced significantly greater GMax EMG activity than the mini squat ($p < 0.05$).
h. The deadlift produced significantly greater GMax EMG activity than the weighted squat through to mini squat ($p < 0.05$).
j. The single-leg vertical jump produced significantly greater GMed EMG activity than the step down through to the mini squat ($p < 0.05$).
k. The single-leg squat and single-leg bridge produced significantly greater GMed EMG activity than the bridge through to the mini squat ($p < 0.05$).
m. The hip hitch produced significantly greater GMed EMG activity than the deadlift through to the mini squat ($p < 0.05$).
n. The side-lying hip abduction produced significantly greater GMed EMG activity than the weighted squat through to the mini squat ($p < 0.05$).
p. These exercises produced significantly greater GMed EMG activity than the mini squat ($p < 0.05$).
q. The lunge produced significantly greater GMed EMG activity than the full squat ($p = 0.023$) mini squat ($p < 0.001$).
r. The counter movement jump produced significantly greater BF EMG activity than the deadlift ($p = 0.025$), weighted squat ($p < 0.001$), single-leg squat ($p = 0.024$), lunge ($p < 0.001$) and step down through to mini squat ($p < 0.05$).
s. The single-leg vertical jump produced significantly greater BF EMG activity than the single-leg squat ($p = 0.016$), lunge ($p < 0.001$) and step down through to mini squat ($p < 0.05$).
t. The single-leg bridge produced significantly greater BF EMG activity than the step down through to mini squat ($p < 0.05$).
u. The deadlift produced significantly greater BF EMG activity than the lateral step up ($p = 0.031$), step up ($p = 0.039$) and mini squat ($p < 0.001$).
v. The raised bridge produced significantly greater BF EMG activity than the full squat through to the mini squat ($p < 0.05$).
w. These exercises produced significantly greater BF EMG activity than the mini squat ($p < 0.05$).

5.1.1 Gluteus Maximus

For the GMax, the single-leg vertical jump (mean ± standard error, 45.6% MVIC ± 4.9%) showed significantly greater mean EMG activation than the mini squat ($p < 0.001$), full squat ($p < 0.001$), weighted squat ($p < 0.001$), raised bridge ($p < 0.001$), bridge ($p < 0.001$), deadlift ($p < 0.001$), prone hip extension ($p = 0.017$), lunge ($p <$

0.001), single-leg squat ($p = 0.012$), step up ($p < 0.001$), step down ($p = 0.002$) and lateral step up ($p = 0.005$). The mini squat exhibited the lowest GMax activity (5.1% MVIC ± 0.7%) which was significantly lower than all the other exercises ($p < 0.05$).

When comparing the three types of bridging exercises, the single-leg bridge (33.6% MVIC ± 3.9%) produced significantly greater mean EMG activity than the raised bridge ($p < 0.001$) and bridge ($p = 0.001$). With respect to the four squat-based exercises, the single-leg squat (26.2% MVIC ± 2.9%) activated the GMax to a significantly greater extent than the mini squat ($p < 0.001$), full squat ($p < 0.001$) and weighted squat ($p < 0.001$). Furthermore, the weighted squat (11.7% MVIC ± 1.9%) and full squat (9.4% MVIC ± 1.1%) produced significantly greater mean EMG activity than the mini squat ($p = 0.004$ and $p = 0.001$ respectively) for the GMax.

5.1.2 Gluteus Medius

The normalised EMG activity for the GMed also revealed the single-leg vertical jump produced the greatest mean EMG activity (51.6% MVIC ± 4.0%). This exercise produced significantly greater EMG activity than the mini squat ($p < 0.001$), full squat ($p < 0.001$), standing hip abduction ($p < 0.001$), raised bridge ($p < 0.001$), weighted squat ($p < 0.001$), deadlift ($p < 0.001$), lunge ($p < 0.001$), bridge ($p < 0.001$), side-lying hip abduction ($p < 0.001$), step up ($p < 0.001$), lateral step up ($p = 0.003$), counter movement jump ($p < 0.001$) and step down ($p = 0.018$). The top five exercises to activate the GMed (single-leg vertical jump, single-leg squat, single-leg bridge, hip hitch and step down) all involved a single limb stance.

With the three bridging exercises, the single-leg type produced the greatest GMed EMG activity (35.0% MVIC ± 3.9%), whilst the raised bridge produced the lowest EMG activity (9.5% MVIC ± 1.7%). The single-leg bridge mean EMG activity for the GMed were significantly greater than the raised bridge ($p < 0.001$) and bridge ($p = 0.001$). There was no significant difference between the EMG activity for the raised bridge and bridge ($p = 0.071$). During the four squat-based exercises, the single-leg squat (35.9% MVIC ± 4.0%) produced significantly greater EMG activity for the GMed than the other three types ($p = 0.000$ for all comparisons). Similarly to the GMax results, the weighted squat (11.0% MVIC ± 1.2%) and full squat (8.6% MVIC ± 1.3%) showed significantly greater EMG activity than the mini squat for the GMed ($p = 0.005$ and $p = 0.007$ respectively). The side-lying hip abduction mean EMG activity (26.1% MVIC ± 2.1%) was significantly greater than that of the standing hip abduction ($p < 0.001$). There was no significant difference in the mean EMG activity between the step up and lateral step up for the GMed ($p = 1.000$).

5.1.3 Biceps Femoris

With respect to the BF, the counter movement jump produced the greatest mean EMG activity (59.2% MVIC ± 10.2%), which were significantly higher than the mini squat ($p < 0.001$), step up ($p = 0.001$), full squat ($p = 0.003$), lateral step up ($p = 0.001$), step down ($p = 0.005$), lunge ($p < 0.001$), single-leg squat ($p = 0.024$), weighted squat ($p < 0.001$) and deadlift ($p = 0.025$). The mini squat, once again, produced the lowest mean EMG activity (14.2% MVIC ± 3.4%) and this result was significantly different to the other three squat-based exercises for the BF ($p < 0.05$). The single-leg bridge activated the BF to 42.3% MVIC, which appeared greater than the EMG activity for the raised bridge and bridge (36.0% MVIC ± 2.6% and 30.5% MVIC ± 4.0% respectively). However, no significant differences were evident between these three exercises ($p > 0.05$).

Table 9. Normalised mean EMG activity in ranked order for the RF, VMO and VL.

Ranking of muscle activity (highest to lowest)	**RF** Exercise	**RF** EMG activity (% MVIC ± SE)	**VMO** Exercise	**VMO** EMG activity (% MVIC ± SE)	**VL** Exercise	**VL** EMG activity (% MVIC ± SE)
1	Counter Movement Jump[a]	78.6 ± 9.3	Single-leg Vertical Jump[g]	112.8 ± 10.1	Single-leg Vertical Jump[p]	100.9 ± 8.0
2	Single-leg Vertical Jump[a]	70.5 ± 6.3	Counter Movement Jump[h]	94.4 ± 11.3	Counter Movement Jump[q]	91.4 ± 8.5
3	Weighted Squat[b]	45.7 ± 7.6	Lateral Step Up[j]	93.7 ± 10.0	Weighted Squat[r]	80.6 ± 11.6
4	Lunge[c]	39.2 ± 7.2	Single-leg Squat[j]	92.1 ± 10.2	Step Down[r]	75.3 ± 4.7
5	Step Up[c]	39.1 ± 5.1	Step Up[j]	92.1 ± 8.3	Lateral Step Up[r]	75.0 ± 7.5
6	Lateral Step Up[c]	37.7 ± 4.9	Step Down[j]	91.3 ± 9.9	Step Up[r]	70.6 ± 5.9
7	Full Squat[d]	36.5 ± 5.0	Lunge[j]	87.1 ± 11.3	Lunge[s]	68.9 ± 8.7
8	Step Down[c]	32.5 ± 4.5	Weighted Squat[j]	84.1 ± 12.7	Single-leg Squat[r]	68.9 ± 6.8
9	Single-leg Squat[c]	31.9 ± 6.3	Full Squat[j]	61.6 ± 5.5	Full Squat[r]	61.9 ± 5.5
10	Straight Leg Raise[c]	30.0 ± 3.2	Deadlift[k]	57.2 ± 8.9	Deadlift[t]	54.1 ± 8.9
11	Deadlift[e]	23.0 ± 3.1	Mini Squat[m]	29.3 ± 3.3	Mini Squat[t]	34.5 ± 4.1
12	Quadriceps Setting[e]	22.4 ± 2.4	Quadriceps Setting[m]	19.1 ± 1.9	Quadriceps Setting[t]	30.4 ± 2.6
13	Mini Squat[e]	14.9 ± 3.3	Straight Leg Raise	9.0 ± 2.2	Straight Leg Raise[u]	16.4 ± 1.8
14	Single-leg Bridge[f]	4.6 ± 0.7	Single-leg Bridge[n]	7.8 ± 1.8	Single-leg Bridge[v]	7.0 ± 0.8
15	Bridge	2.9 ± 0.5	Bridge	5.3 ± 1.5	Bridge	4.9 ± 1.1
16	Raised Bridge	2.5 ± 0.4	Raised Bridge	4.4 ± 1.0	Raised Bridge	4.4 ± 0.7

RF: Rectus Femoris, VMO: Vastus Medialis Oblique, VL: Vastus Lateralis, MVIC: Maximal Voluntary Isometric Contraction, SE: Standard Error.

a. The counter movement jump and single-leg vertical jump produced significantly greater RF EMG activity than the weighted squat through to the raised bridge ($p < 0.05$).
b. The weighted squat produced significantly greater RF EMG activity than the deadlift through to the raised bridge ($p < 0.05$).
c. These exercises produced significantly greater RF EMG activity than the mini squat through to the raised bridge ($p < 0.05$).
d. The full squat produced significantly greater RF EMG activity than the quadriceps setting through to the raised bridge ($p < 0.05$).
f. The single-leg bridge produced significantly greater RF EMG activity than the bridge ($p = 0.008$) and raised bridge ($p < 0.05$).
g. The single-leg vertical jump produced significantly greater VMO EMG activity than the single-leg squat ($p < 0.001$) and full squat through to the raised bridge ($p < 0.05$).
h. The counter movement jump produced significantly greater VMO EMG activity than the deadlift through to the raised bridge ($p < 0.05$).
j. These exercises produced significantly greater VMO EMG activity than the min squat through to the raised bridge ($p < 0.05$).
k. The deadlift produced significantly greater VMO EMG activity than the quadriceps setting through to the raised bridge ($p < 0.05$).
m. The quadriceps setting and mini squat produced significantly greater VMO EMG activity than the straight leg raise through to the raised bridge ($p < 0.05$).
n. The single-leg bridge produced significantly greater VMO EMG activity than the bridge ($p = 0.038$) and raised bridge ($p = 0.034$).
p. The single-leg vertical jump produced significantly greater VL EMG activity than the single-leg squat ($p = 0.003$) and the deadlift through to the raised bridge ($p < 0.05$).
q. The counter movement jump produced significantly greater VL EMG activity than the lunge ($p = 0.026$) and the deadlift through to the raised bridge ($p < 0.05$).
r. These exercises produced significantly greater VL EMG activity than the mini squat through to the raised bridge ($p < 0.05$).
s. The lunge produced significantly greater VL EMG activity than the quadriceps setting through to the raised bridge ($p < 0.05$).
t. These exercises produced significantly greater VL EMG activity than the straight leg raise through to the raised bridge ($p < 0.05$).
u. The straight leg raise produced significantly greater VL EMG activity than the single-leg bridge through to the raised bridge ($p < 0.05$).
v. The single-leg bridge produced significantly greater VL EMG activity than the raised bridge ($p = 0.001$).

5.1.4 Rectus Femoris

The counter movement jump produced the greatest mean EMG activity for the RF (78.6% MVIC ± 9.3%). This result was significantly greater than the EMG activity for all the other exercises for this muscle ($p < 0.05$), excluding the single-leg vertical jump. The three types of bridges produced the least RF EMG activity, with these results being significantly lower than all of the other exercises ($p < 0.05$). The non weight bearing straight leg raise exercise (30.0% MVIC ± 3.2%) produced significantly greater mean EMG activity than the mini squat for the RF ($p < 0.001$). When considering the squat-based exercises, the mini squat mean EMG activity was significantly lower than the mean EMG activity for the full squat, weighted squat and single-leg squat ($p < 0.05$). There were no significant differences evident for the mean EMG activity between the latter three types of squat ($p = 1.000$ for all comparisons) or between the step up, step down and lateral step up for the RF ($p = 1.000$ for all comparisons).

5.1.5 Vastus Medialis Oblique

For the VMO, the single-leg vertical jump produced the greatest mean EMG activity (112.8% MVIC ± 10.1%). This EMG value was significantly greater than the three bridging exercises ($p < 0.001$), the straight leg raise ($p < 0.001$), quadriceps setting ($p < 0.001$), mini squat ($p < 0.001$), deadlift ($p = 0.001$), full squat ($p < 0.001$) and single-leg squat ($p = 0.014$). For the two non weight bearing rehabilitation exercises, the quadriceps setting activated the VMO to 19.1% MVIC, which was a significantly greater EMG activity than produced by the straight leg raise ($p = 0.044$). The mini squat produced lower mean EMG activity for the VMO than the weighted squat ($p < 0.001$), single-leg squat ($p < 0.001$) and full squat ($p < 0.001$). No significant differences were evident between mean EMG activation levels for the VMO during the step up, step down and lateral step up ($p < 0.001$ for all comparisons).

5.1.6 Vastus Lateralis

Similarly to the VMO, the single-leg vertical jump produced the greatest mean EMG activity for the VL (100.9% MVIC ± 8.0%). This result for the single-leg vertical jump was significantly greater than the mean EMG activity for the three bridges ($p < 0.001$), straight leg raise ($p < 0.001$), quadriceps setting ($p < 0.001$), mini squat ($p < 0.001$), deadlift ($p < 0.001$) and single-leg squat ($p = 0.003$). As with the results for the other quadriceps, the bridges produced the lowest mean EMG activity for the VL. The straight leg raise (16.4% MVIC ± 1.8%) produced significantly lower mean EMG activity than the quadriceps setting exercise ($p = 0.003$). For the VL, the mini squat mean EMG activity (34.6% MVIC ± 4.1%) was significantly lower than the full squat ($p < 0.001$), single-leg squat ($p = 0.037$) and weighted squat ($p = 0.003$). There were no significant differences evident between the mean EMG activity for the step up, step down and lateral step up ($p = 1.000$ for all comparisons).

5.2 Muscle Ratios

5.2.1 Vastus Medialis Oblique:Vastus Lateralis Ratio

The results for the VMO:VL ratio (Figure 4) highlighted that the straight leg raise, quadriceps setting, mini squat and counter movement jump failed to preferentially activate the VMO in respect to the VL. All other exercises produced a ratio value in excess of one, which meant the VMO was activated to a greater extent than the VL.

The single-leg squat produced the highest VMO:VL ratio (1.66 ± 0.41), whilst the straight leg raise created the lowest ratio (0.44 ± 0.10). The straight leg raise VMO:VL ratio was significantly lower than the ratio results for all of the other exercises ($p < 0.05$) excluding the weighted squat. Furthermore, the quadriceps setting exercise produced a significantly lower VMO:VL ratio than the mini squat ($p = 0.016$), full squat ($p < 0.001$), step up ($p < 0.001$), step down ($p = 0.005$), lateral step up ($p = 0.004$), single-leg squat ($p = 0.044$), counter movement jump ($p = 0.026$) and single-leg vertical jump ($p < 0.001$).

Due to the three bridging exercises focussing specifically on the posterior musculature, shown by the quadriceps mean EMG activity never exceeding 8% MVIC, these three exercises were not included in the VMO:VL ratio analysis.

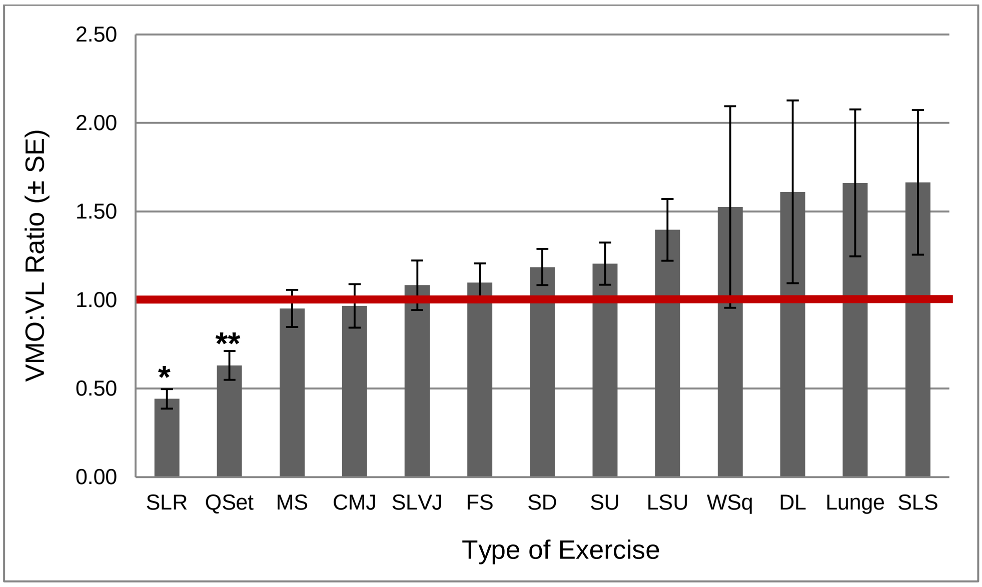

Figure 4. The continuum of exercises to produce an increasing VMO:VL ratio.

SE: Standard Error, VMO: Vastus Medialis Oblique, VL: Vastus Lateralis, SLR: Straight Leg Raise, QSet: Quadriceps Setting, MS: Mini Squat, CMJ: Counter Movement Jump, SLVJ: Single-leg Vertical Jump, FS: Full Squat, SD: Step Down, SU: Step Up, LSU: Lateral Step Up, WSq: Weighted Squat, DL: Deadlift, SLS: Single-leg Squat.

** The SLR produced a significantly lower VMO:VL ratio than all the other exercises ($p < 0.05$) excluding the WSq ($p = 0.095$).*

*** The QSet produced a significantly lower VMO:VL ratio than the MS ($p = 0.016$), CMJ ($p = 0.026$), SLVJ ($p < 0.001$), FS ($p < 0.001$), SD ($p = 0.005$), SU ($p < 0.001$), LSU ($p = 0.004$) and SLS ($p = 0.044$).*

5.2.2 Hamstrings:Quadriceps Ratio

The results (Figure 5) showed the bridge, single-leg bridge, raised bridge and deadlift were the only exercises to produce a HS:Quads ratio in excess of one, which meant that these exercises activated the hamstrings to a greater extent than the quadriceps.

The raised bridge produced the overall highest ratio (16.61 ± 3.54). The three bridging exercises produced a significantly higher HS:Quads ratio than all of the other exercises ($p < 0.05$). Between the three bridges, the raised bridge produced a HS:Quads ratio, which was significantly greater than the single-leg bridge ($p = 0.038$). The deadlift produced a significantly higher ratio than the weighted squat ($p = 0.001$) and mini squat ($p = 0.010$), but apart from this, there were no other differences between the HS:Quads ratios for the remaining exercises ($p > 0.05$).

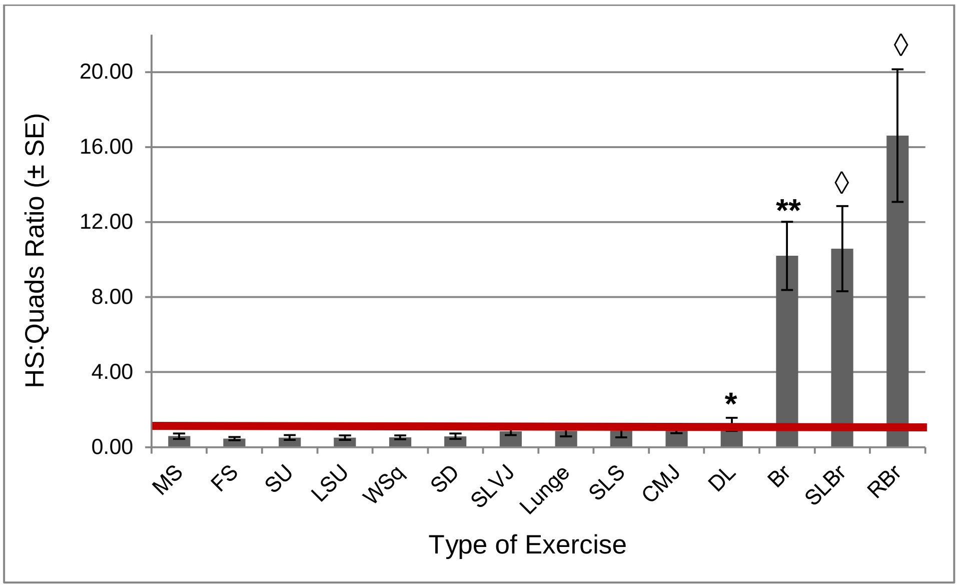

Figure 5. The continuum of exercises to produce an increasing HS:Quads ratio.

SE: Standard Error, HS: Hamstrings (Biceps Femoris), Quads: Quadriceps (Rectus Femoris, Vastus Medialis Oblique, Vastus Lateralis), MS: Mini Squat, FS: Full Squat, SU: Step Up, LSU: Lateral Step Up, WSq: Weighted Squat, SD: Step Down, SLVJ: Single-leg Vertical Jump, SLS: Single-leg Squat, CMJ: Counter Movement Jump, DL: Deadlift, Br: Bridge, SLBr: Single-leg Bridge, RBr: Raised Bridge.

* *The DL produced a significantly greater HS:Quads ratio than the MS (p = 0.019) and WSq (p = 0.043).*

** *The Br produced a significantly greater HS:Quads ratio than all the exercises (p < 0.05) excluding the SLBr and RBr.*

◊ *The SLBr and RBr produced significantly greater HS:Quads ratio than all the exercises (p < 0.05) excluding the Br.*

5.2.3 Gluteus Maximus:Biceps Femoris Ratio

With regards to the GMax:BF ratio (Figure 6), the step up produced the highest value (1.55 ± 0.88), whilst the raised bridge produced the lowest ratio (0.38 ± 0.27). The prone hip extension, single-leg squat, single-leg vertical jump, step down, lateral step up and step up preferentially activated the GMax over the HS, shown by the ratios for these exercises exceeding the value of one. The step up, step down and lateral step up produced significantly greater ratios than the mini squat, full squat and raised bridge ($p < 0.05$). Both the step up and lateral step up also produced significantly higher GMax:BF ratios than the weighted squat and deadlift ($p < 0.05$). The raised bridge produced a significantly lower ratio than the single-leg squat (p = 0003), single-leg vertical jump (p = 0.005) and single-leg bridge (p = 0.012). Finally, the weighted squat produced a significantly lower GMax:BF ratio than the single-leg squat (p = 0.020), step down (p = 0.005) and deadlift (p = 0.007).

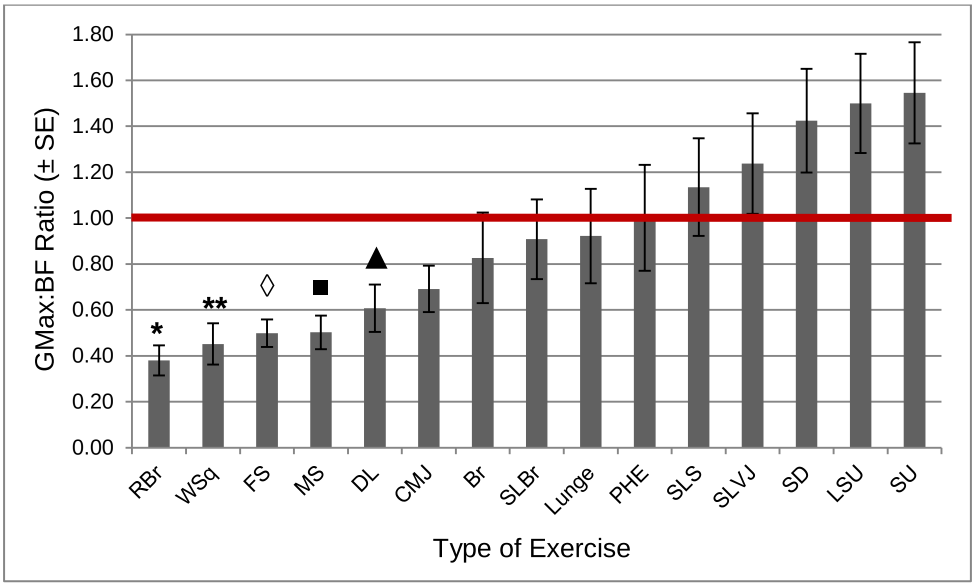

Figure 6. The continuum of exercises to produce an increasing GMax:BF ratio.

SE: Standard Error, GMax: Gluteus Maximus, BF: Biceps Femoris, RBr: Raised Bridge, WSq: Weighted Squat, FS: Full Squat, MS: Mini Squat, DL: Deadlift, CMJ: Counter Movement Jump, Br: Bridge, SLBr: Single-leg Bridge, PHE: Prone Hip Extension, SLS: Single-leg Squat, SLVJ: Single-leg Vertical Jump, SD: Step Down, LSU: Lateral Step Up, SU: Step Up.

** The RBr produced a significantly lower GMax:BF ratio than the SLBr (p = 0.012), SLS (p = 0.003), SLVJ (p = 0.005), SD (p = 0.001), LSU (p < 0.001) and SU (p = 0.003).*

*** The WSq produced a significantly lower GMax:BF ratio than the DL (p = 0.007), SLS (p = 0.020), SLVJ (p = 0.010), SD (p = 0.005), LSU (p = 0.001) and SU (p < 0.001).*

◊ The FS produced a significantly lower GMax:BF ratio than the SD (p < 0.001), LSU (p = 0.001) and SU (p = 0.008).

■ The MS produced a significantly lower GMax:BF ratio than the SD (p = 0.025), LSU (p = 0.006) and SU (p = 0.005).

▲ The DL produced a significantly lower GMax:BF ratio than the LSU (p = 0.026) and SU (p = 0.012).

5.3 Gender Differences

The results of the independent t-tests comparing the mean EMG activity between males and females are shown in Table 10. Females had a significantly greater GMax EMG activation compared to males during the raised bridge (p = 0.044), lunge (p = 0.003), step down (p = 0.02), counter movement jump (p = 0.022) and single-leg vertical jump (p = 0.003). The mean EMG activity for the GMed was also significantly greater during the lunge for the females compared to the males (p = 0.031). There were no significant differences between males and females for EMG muscle activity of the GMax and GMed for the remaining exercises (p > 0.05).

Males had a significantly greater BF EMG activation during the single-leg vertical jump compared to the females (p = 0.036) but no other differences were evident between genders for this muscle (p > 0.05). During the step up (p = 0.038), step down (p = 0.011) and lateral step up (p = 0.024), the females' mean EMG activity for the RF were significantly greater than for the males. Finally, the females produced a significantly greater VMO EMG activation during the straight leg raise (p = 0.045) but there were no further significant differences for the VMO or VL EMG activation when comparing between the two genders.

Table 10. The normalised mean EMG activity for males and females during the rehabilitation exercises.

Exercise	**Muscle**	**Mean EMG %MVIC**		**p value**
		Males	**Females**	
Quadriceps Setting	RF	23.15	21.52	0.879
	VMO	19.22	18.86	0.696
	VL	32.60	28.21	0.578
Straight Leg Raise	RF	26.17	33.39	0.244
	VMO	5.32	12.73	0.045
	VL	14.39	18.60	0.163
Side-lying Hip Abduction	GMed	28.33	23.84	0.261
Standing Hip Abduction	GMed	7.10	11.38	0.461
Prone Hip Extension	GMax	22.56	17.33	0.725
	BF	30.80	28.21	0.731
Mini Squat	GMax	5.33	4.86	0.891
	GMed	5.12	5.45	0.921
	BF	18.47	9.88	0.169
	RF	11.34	18.39	0.218
	VMO	23.47	35.86	0.058
	VL	31.82	37.28	0.779
Full Squat	GMax	8.63	10.16	0.363
	GMed	6.44	10.83	0.147
	BF	23.23	20.57	0.449

	RF	36.78	36.19	0.691
	VMO	59.27	64.99	0.358
	VL	65.72	56.96	0.473
Lunge	GMax	12.16	27.94	0.003
	GMed	12.47	21.06	0.031
	BF	40.06	24.37	0.1
	RF	30.53	47.89	0.231
	VMO	71.96	104.05	0.148
	VL	58.47	79.33	0.222
Bridge	GMax	15.56	17.23	0.277
	GMed	14.91	18.75	0.727
	BF	27.06	33.87	0.389
Single-leg Bridge	GMax	27.15	39.26	0.105
	GMed	35.99	34.08	0.917
	BF	43.50	40.92	0.74
Raised Bridge	GMax	8.33	16.04	0.044
	GMed	10.64	8.23	0.575
	BF	34.97	37.02	0.973
Step Up	GMax	20.75	33.07	0.076
	GMed	27.89	26.73	0.853
	BF	19.66	22.92	0.831
	RF	29.86	49.47	0.038
	VMO	77.35	106.78	0.075
	VL	65.69	78.06	0.362
Step Down	GMax	18.48	38.57	0.02
	GMed	31.43	32.85	0.663
	BF	27.19	23.43	0.522
	RF	22.49	41.31	0.011
	VMO	74.73	110.26	0.051
	VL	70.60	81.36	0.206
Lateral Step Up	GMax	22.48	36.21	0.08
	GMed	30.69	30.19	0.958
	BF	24.30	23.15	0.653
	RF	27.63	47.67	0.024
	VMO	83.92	108.37	0.174

	VL	64.60	88.38	0.147
Hip Hitch	GMed	31.40	37.92	0.693
Single-leg Squat	GMax	22.11	30.36	0.123
	GMed	38.35	33.97	0.735
	BF	36.46	30.32	0.317
	RF	30.42	33.26	0.935
	VMO	77.07	107.15	0.162
	VL	73.91	62.40	0.889
Counter Movement Jump	GMax	25.31	46.97	0.022
	GMed	26.88	36.43	0.086
	BF	75.05	38.84	0.058
	RF	82.86	73.72	0.703
	VMO	80.47	110.32	0.111
	VL	80.61	103.73	0.167
Single-leg Vertical Jump	GMax	32.64	58.54	0.003
	GMed	46.98	56.77	0.261
	BF	62.97	37.11	0.036
	RF	72.77	68.14	0.473
	VMO	109.04	118.88	0.7
	VL	96.70	105.07	0.512
Weighted Squat	GMax	10.52	14.08	0.485
	GMed	11.34	10.45	0.807
	BF	42.40	21.92	0.088
	RF	49.83	37.54	0.619
	VMO	76.64	98.96	0.391
	VL	76.76	88.33	0.685
Deadlift	GMax	14.97	23.16	0.093
	GMed	12.98	19.61	0.319
	BF	45.91	30.61	0.417
	RF	17.70	29.28	0.073
	VMO	56.24	58.74	0.724
	VL	46.86	62.57	0.488

GMax: Gluteus Maximus, GMed: Gluteus Medius, BF: Biceps Femoris, RF: Rectus Femoris, VMO: Vastus Medialis Oblique, VL: Vastus Lateralis.

Significance was determined as $p < 0.05$ (significant values highlighted in the table).

5.4 Knee Flexion Angles

The mean knee flexion angle during the initial descent for the counter movement jump was significantly greater than the knee flexion angle during the same stage of the single-leg vertical jump ($p < 0.001$) (Figure 7). However, there were no significant differences between the amount of knee flexion exhibited by the males compared to the females during the counter movement phase of the counter movement jump ($p = 0.810$) or the single-leg vertical jump ($p = 0.119$) (Figure 8).

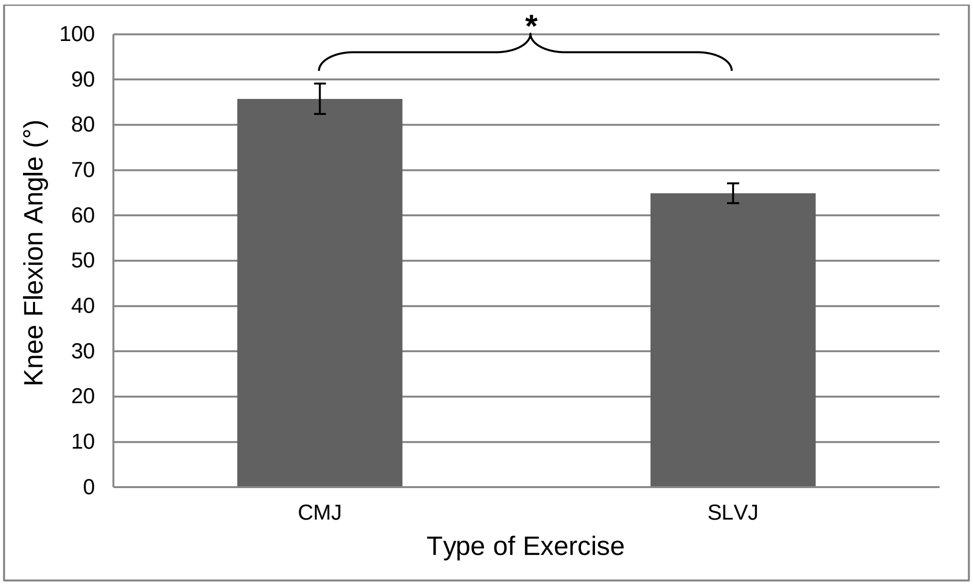

Figure 7. The mean knee flexion angles during the initial descent phase of the counter movement jump and single-leg vertical jump.

CMJ: Counter Movement Jump, SLVJ: Single-leg Vertical Jump.
**The knee flexion angle during the initial descent phase of the CMJ was a significantly greater angle than observed during the SLVJ (p < 0.001).*

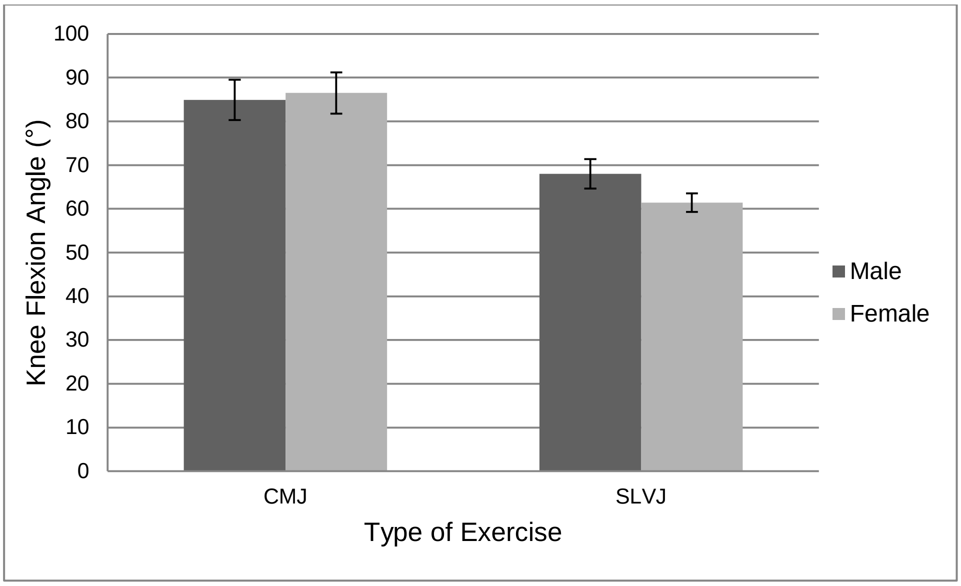

Figure 8. The mean knee flexion angles for each gender during the initial descent phase of the counter movement jump and single-leg vertical jump.

CMJ: Counter Movement Jump, SLVJ: Single-leg Vertical Jump.
There were no significant differences for the knee flexion angles between the genders for either exercise (p > 0.05).

There were also no significant differences between the knee flexion angles of the males and females during the step up, step down and lateral step up ($p > 0.05$) (Figure 9).

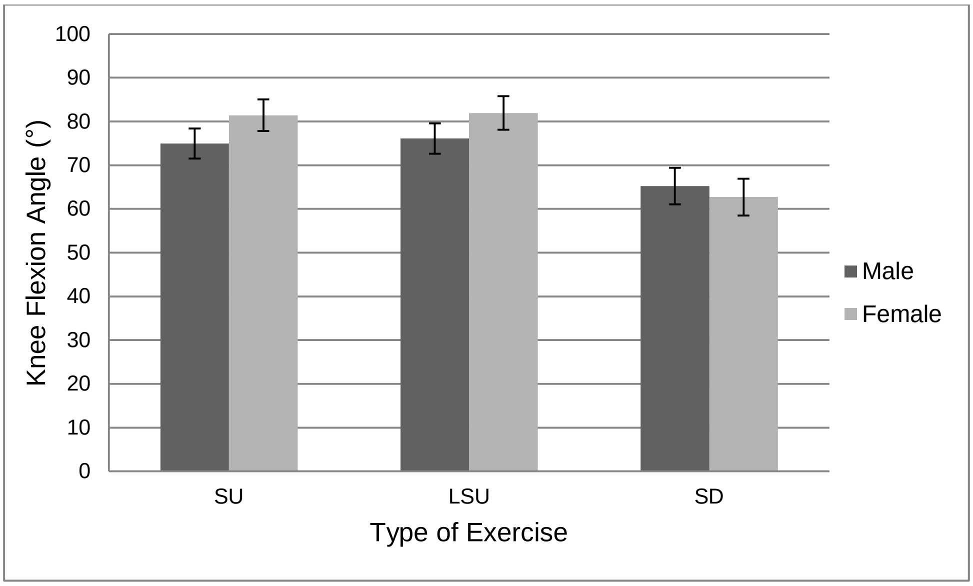

Figure 9. The mean knee flexion angles for males and females during the step up, lateral step up and step down.

SU: Step Up, LSU: Lateral Step Up, SD: Step Down.
No significant differences were evident between the genders for the knee flexion angle during the step up, lateral step up and step down (p > 0.05).

Chapter Six: Discussion

This study was conducted to quantify hip and knee muscle activity using EMG during twenty lower limb rehabilitation exercises, in order to identify a continuum of exercises in relation to the amount of muscle activation from the GMax, GMed, BF, RF, VMO and VL. The greatest mean EMG activity for each muscle was produced by either the single-leg vertical jump or the counter movement jump, both explosive exercises. These exercises were within the 'performance enhancing' section as they are used to increase power and cannot be performed until the athlete has full lower limb function. They are, therefore, likely exercises to be at the top end of the continuum. Jakobsen et al. (2013) identified that ballistic exercises had the ability to produce similar or even larger EMG activity than high-load, slow tempo exercises, potentially due to the type of muscle fibre recruitment. The current results followed a similar pattern with the two jump exercises producing the greatest EMG activity compared with high-load exercises such as the weighted squat and deadlift. For the GMax, GMed, BF and RF, the mini squat exercise produced the lowest mean EMG activity. This exercise was partial weight bearing so was classified as an 'intermediate' exercise. However, it produced less EMG activity for these aforementioned muscles than the non weight bearing 'early' exercises. The mean EMG activity for each muscle will be discussed in Section 6.1.

A secondary aim was to analyse the VMO:VL ratio, HS:Quads ratio and GMax:BF ratio during the same exercises. These will be discussed in Section 6.2 (page 91). The additional aim of determining the effect of gender on hip and knee muscle activity during the rehabilitation exercises will be discussed in Section 6.3 (page 98).

6.1 Normalised EMG Activity

6.1.1 Gluteus Maximus

The prone hip extension was non weight bearing but produced significantly greater mean EMG activity for the GMax than the mini squat, a partial weight bearing exercise. Previous research for prone hip extension has been more focussed on the onset timing of the GMax, hamstrings and erector spinae rather than on the EMG magnitudes (Bruno et al., 2008; Chance-Larsen et al., 2010; Lehman et al., 2004). Therefore, there were limited studies to compare the current results to. However, it can be proposed that the GMax produced greater EMG activity during the prone hip extension as opposed to the mini squat because the non weight bearing exercise involved hip extension against gravity and also the leg had a greater lever length (Bolgla & Uhl, 2005). Some clinicians used the prone hip extension exercise to replicate the muscles' function to extend the hip from a standing position (Bruno et al., 2008; Lehman et al., 2004; Sakamoto et al., 2009). The current results showed this movement was only able to activate the GMax to 20% MVIC. As the pelvis was stabilised by the floor, the test may not have reflected functional activity so the purpose of the test should be viewed with caution.

Within the 'intermediate' stage exercises, the mini squat and full squat produced significantly less mean EMG activity than the lunge. The mini squat only involved 45° of knee flexion whereas the full squat and lunge lowered the dominant knee to 90° of knee flexion. Jakobsen et al. (2013) concluded that the GMax, VMO and VL EMG activity increased as the knee flexed to a greater angle during the lunge, hence the greater EMG activity in this study for the lunge compared to the mini squat. However, as the full squat and lunge incorporated the same amount of knee flexion, the difference in EMG activity between these two exercises may have been due the positioning of the centre of mass in relation to the front foot (dominant side) during the lunge (Farrokhi et al., 2008). It was previously observed that ensuring a vertical shank position, as seen with the lunge, increased the mechanical demands on the hip extensor net joint moment (Mathiyakom et al., 2005). This would also be comparable to the findings of Ayotte et al. (2007) who concluded that a wall squat with the foot placed anteriorly of the hip produced greater GMax EMG activity than a traditional full squat. To progress with the rehabilitation and activate the GMax to a greater extent, these results suggest the knee should be flexed to a larger angle, and the centre of mass should be posterior to the dominant foot during double-leg CKC exercises.

The mini squat mean EMG activity for the GMax was also significantly lower than all three types of bridging exercises. During the bridges, the GMax had to extend the hip against gravity from the supine position, which may have been the reasoning for activating the muscle to a greater extent than the mini squat (Ishida et al., 2011). Between the bridging exercises, the single-leg bridge produced significantly greater EMG activity than the raised bridge and bridge. This means that during rehabilitation for the GMax, it would be recommended to complete the raised bridge, followed by the bridge, then single-leg bridge to continually progress the muscle. During the single-leg bridge, the GMax was used to extend the hip, from a flexed position to a neutral position, as well as to support the pelvis. Ekstrom et al. (2007) reported a value of 40.0% MVIC for the GMax during a single-leg bridge whilst Selkowitz et al. (2013) concluded a result of 34.6% MVIC. Both of these were similar to the findings in the current study (33.6% MVIC). However, Boren et al. (2011) reported a higher amount of GMax activation during the single-leg bridge (54.2% MVIC). During this exercise, any difference in the pelvic tilt may have affected the results because it was identified that a posterior pelvic tilt significantly increased the GMax EMG activation in a study by Ishida et al. (2011) due to the position activating the hip extensors.

For the 'late' stage rehabilitation exercises, no significant differences were evident for the mean EMG activity for the GMax so it can be proposed that these exercises can be used at the same stage of rehabilitation. All of these exercises involved a single-leg stance so the GMax had to resist internal rotation of the femur as well as extend the hip. During the step up, lateral step up and step down, the trunk was held in an upright position throughout and ensured a forceful hip extension thrust. The current study recorded a value of 29.8% MVIC for the GMax during the lateral step up, which was comparable to 29.0% MVIC, which was identified by Ekstrom et al. (2007). However, these results differed from those of Ayotte et al. (2007) (56.0% MVIC) and Boren et al. (2011) (63.8% MVIC). The MVIC for the GMax was performed in a supine position with 30° of hip flexion in the study by Ayotte et al. (2007) as opposed to the more standard prone position in this study, which could have affected the results. In addition, Boren et al. (2011) used a 15 cm step for the lateral step up in contrast to the 30 cm step used in the current study. As their EMG activity was greater than this study, it may be hypothesised that their data analysis process of only analysing the 100 ms surrounding the peak EMG value had a greater impact on the results, rather than the step height. In a more functional sense, the higher activation of the GMax in the current results during a step up were comparable to the results of those which investigated uphill running (Swanson & Caldwell, 2000) and stair ascent (Lyons et al., 1983). During uphill running, the hip extension from an increased hip flexion position activated the GMax to a greater extent than level running (Swanson & Caldwell, 2000). This is an important concept for clinicians to be aware of when prescribing rehabilitation exercises for the GMax.

The 'performance enhancing' weighted squat produced significantly less mean EMG activity than the deadlift, single-leg bridge, both types of jump exercises and all 'late' stage exercises, excluding the hip hitch. The deadlift also produced significantly less EMG activity than the single-leg bridge and two jumps. During a deadlift, the GMax works together with the erector spinae and hamstrings to control the trunk flexion at the sacroiliac joint and lumbar vertebrae (Lieberman et al., 2006). The deadlift may have produced greater GMax results than the weighted squat due to a higher focus on the 'upward' phase during the deadlift which required forceful hip extension (Hales et al., 2009). Potentially, if the barbell was loaded to a greater extent, this exercise would have produced greater GMax EMG activity than the 18% MVIC recorded. This result for the deadlift appeared lower than identified in other studies which ranged between 35-59% MVIC (Distefano et al., 2009; Escamilla et al., 2002). During deeper knee flexion angles, the GMax was previously found to produce greater EMG activity, possibly due to having to decelerate the bar at the end of the lift (Escamilla et al., 2002). The current study examined the bent leg deadlift as opposed to the stiff leg alternative so this could be another reason for varied values when comparing the mean EMG activity (Wright et al., 1999).

When considering the four types of squats, the mini squat produced significantly less mean EMG activity for the GMax than the full squat, weighted squat and single-leg squat. The full squat and weighted squat also produced significantly less mean EMG activity than the single-leg squat. The effect of different squat depths on the GMax EMG activation has been previously investigated (Caterisano et al., 2002). The current results showed that a squat to 90° of knee flexion produced significantly greater mean EMG activity than a squat to 45°, so it could be speculated that deeper squats were more effective for strengthening the GMax. These results would therefore recommend that deeper squats should be used as a progression during rehabilitation for the GMax. Caterisano et al. (2002) also reported that the deeper the squat range of motion, the greater the GMax EMG activity. Having said this, the current results showed that out of the four squats, the single-leg squat produced the greatest mean EMG activity for the GMax irrespective of this exercise only lowering to 60° of knee flexion. This may have been due to the increased difficulty of the exercise by requiring the body to be stabilised over a smaller base of support. The GMax externally rotates the femur so during the single-leg squat, this muscle would have also been activating to resist femoral internal rotation (Ling & Kumar, 2006). For the single-leg squat, the present study and Boudreau et al. (2009) identified comparable results for the GMax (26.2% MVIC and 35.2% MVIC respectively). This was not a standard figure throughout all the literature, though, with other studies concluding values as high as 70.2% MVIC (Boren et al., 2011; Lubahn et al., 2011). The current study used a standardised knee flexion angle of 60°, but others have chosen to use a chair for subjects to lower their buttocks to, which meant taller subjects would have been squatting to a larger extent than shorter subjects (Boren et al., 2011). This could have led to variations within the results. Finally, an additional 40% of body mass resistance during a squat to 90° of knee flexion was unable to increase the GMax activation. Overall, the current results highlighted that a deeper squat range of motion, consequently increased the GMax EMG activation. However, when needing to activate the GMax to an even greater extent, a single-limb stance should be incorporated due to this muscle providing pelvic stability and resisting internal rotation of the femur, in addition to its main role of extending the hip.

No significant differences were evident between the counter movement jump and single-leg vertical jump for the mean EMG activity of the GMax, although on the continuum, the single-leg vertical jump produced the greatest EMG activity. During both of these movements there was a forceful triple extension of the hip, knee and ankle. The GMax was recruited to extend the hip during the upward phase, as well as to work eccentrically to control the hip flexion upon landing. During the single-leg vertical jump, the GMax had the additional role of preventing internal rotation of the femur as highlighted before. The single-leg vertical jump was the only exercise for the GMax to surpass the 40% threshold which is thought to be needed to strengthen the muscle (Ayotte et al., 2007; Escamilla et al., 2010). Therefore, it is an appropriate rehabilitation, prehabilitation and performance enhancing exercise in situations where the GMax needs to be strengthened.

6.1.2 Gluteus Medius

The mini squat also produced the lowest EMG activity for the GMed at a value of only 5.3% MVIC. This was a significantly lower EMG activation than all the other exercises excluding the standing hip abduction and raised bridge. The EMG activity results for the GMed in the study by Lubahn et al. (2011) were greater during the mini squat (17.6% MVIC) but this may have been due to them using the peak EMG rather than mean values during the data analysis. The mini squat, along with the full squat, standing hip abduction and raised bridge appeared to have minimal training effects for the GMed due to the mean EMG activity being less than 10% (Escamilla et al., 2010). These exercises are therefore not recommended when needing to strengthen this muscle.

The side-lying hip abduction produced significantly greater mean EMG activity than the standing hip abduction for the GMed; this was a similar finding to those of Bolgla and Uhl (2005). During the two hip abduction exercises, the GMed activated to overcome the external torque produced by the length of the lever and the mass of the leg (Nordin & Frankel, 2001). The mass of the leg was identified as being approximately 16% of the total body mass (Bolgla & Uhl, 2005). The EMG activity may have been higher during the side-lying hip abduction exercise as opposed to the standing hip abduction exercise due to a greater effect from gravity during the side-lying method (Bolgla & Uhl, 2005). The GMed activated to 26.1% MVIC during the side-lying hip abduction, but there was no consensus for the average value within the literature, with results ranging from 39-81% MVIC (Bolgla & Uhl, 2005; Boren et al., 2011; Distefano et al., 2009; Ekstrom et al., 2007; Selkowitz et al., 2013). These discrepancies were likely to be due to methodological differences including the use of an ankle weight, the range of motion and the position of the non-dominant leg with regards to how much the pelvis was supported. From the current results, the side-lying hip abduction should be performed rather than the standing alternative when needing to activate the GMed during rehabilitation.

During single-leg stance, the hip abductors, including the GMed, are required to stabilise the pelvis in the frontal plane (Neumann & Hase, 1994). This role is achieved by preventing adduction of the contralateral hip, which means providing enough muscular activity to support the head, trunk, arms and non weight bearing leg; these have been stated to comprise 84% of a person's body mass (Bolgla & Uhl, 2005). This was likely to be the reasoning behind the top five exercises for the GMed only having the dominant foot in contact with the floor. This concept was in agreement with the results of a study by Youdas et al. (2013), who concluded the GMed was activated significantly more on the stance leg, rather than the leg performing hip abduction, during a lateral band walk. For the strengthening of the GMed, it is recommended that this concept is understood and applied to the rehabilitation once the athlete is able to fully weight bear.

The single-leg squat produced mean EMG activity four times greater than the double-leg alternative, which highlighted the importance of a single-leg stance for activating the GMed. In the current study, the single-leg squat produced results of 35.9% MVIC for the GMed, which was similar to the values found by Ayotte et al. (2007) (36.0% MVIC) and Boudreau et al. (2009) (30.1% MVIC) but lower than those of Boren et al. (2011) (82.3% MVIC) and Lubahn et al. (2011) (47.5% MVIC). The latter two studies used peak EMG readings as stated before which could have led to larger normalised EMG activity. Furthermore, Lubahn et al. (2011) allowed subjects to hold onto a pole for balance but this could have affected the results by them tilting their trunk and shifting their mass within the base of support.

No differences were evident between the 'late' stage exercises, all of which measured the GMed on the weight bearing single-leg. In a clinical setting, the step down, single-leg squat, hip hitch and both step ups (forwards and lateral) can all be used at the same stage of rehabilitation due to the apparent similar GMed activation levels. Ayotte et al. (2007) also observed no significant differences between the step up and lateral step up. The lateral step up incorporated a sideways movement, which could have activated the hip abductors to a greater extent as the weight transferred laterally. However, within the closed setting of this particular exercise, this reasoning was not justified with no significant differences in the activation of the GMed during the front and lateral step ups. The mean EMG activation for the GMed during the step up and hip hitch in the current study were remarkably similar to the results by Selkowitz et al. (2013) (step up: 27.3% MVIC and 29.5% MVIC respectively; hip hitch: 34.7% MVIC and 37.7% MVIC respectively). During the hip hitch, the hip abductors work eccentrically to adduct the contralateral hip, then concentrically to abduct the hip so the pelvis become level again (Bolgla & Uhl, 2005). The 'late' stage exercises produced greater EMG activity than the deadlift and weighted squat, most likely due to the latter two exercises being double-legged so the wider stance supported the pelvis and upper body with less muscular activity requirement. Therefore, for the purpose of wanting to strengthen the GMed, it can be suggested to use body weight exercises with a single-leg stance, rather than weighted double-leg exercises.

The single-leg vertical jump produced significantly greater mean EMG activity than the counter movement jump, and all the other exercises excluding the hip hitch, single-leg bridge and single-leg squat. The single-leg vertical jump was highly dynamic, causing the GMed to activate to stabilise the pelvis and upper body. The GMed is an important muscle to train during this explosive situation to avoid femoral adduction and knee valgus forces, both of which have been proposed as factors which increase the likelihood of damaging the ACL (Chaudhari & Andriacchi, 2006; Lawrence et al., 2008). However, if unable to fully weight bear on the injured side, it is worth noting that the single-leg bridge did not produce significantly different GMed EMG activity from the single-leg vertical jump. It can therefore be speculated that the single-leg bridge could be used to strengthen the GMed earlier on in rehabilitation but with similar activation levels. Previous literature has investigated the GMed activation during a single-leg bridge and obtained values of 47.0% MVIC (Ekstrom et al., 2007) and 55.0% MVIC (Boren et al., 2011). These were slightly higher results than the value of 35.0% MIVC which was reported in the current study. However, the trend in the present study whereby the single-leg bridge recruited the GMed to a greater extent than the bridge was in agreement to the observations by Ekstrom et al. (2007).

6.1.3 Biceps Femoris

The mean EMG activity for the BF during the 'early' stage prone hip extension was not significantly different to the EMG activity for any of the other exercises. This differed from the GMax results, whereby this non weight bearing exercise produced significantly greater mean EMG activity than the mini squat and significantly less activity than the single-leg vertical jump. For the BF, the prone hip extension activated the muscle to 30% MVIC so it can be proposed that this exercise is used for initial rehabilitation and endurance purposes with high repetitions before the patient is able to weight bear (Jacobs et al., 2009).

Within the 'intermediate' exercise section, the mini squat produced significantly less mean EMG activity than the full squat for the BF, showing the knee flexion angle affected muscular activity. Previous research by Caterisano et al. (2002) concluded there were no differences in BF activity during different depths of squats. However, their subjects performed squats with a barbell containing 100-125% of body mass as additional resistance, so during this heavily weighted condition, knee range of motion may have had less of an impact on BF activity. A small sample size of 10 was also used so their results should be viewed with caution. For the mini squat and full squat, the current results showed the BF was activated to 14.2% MVIC and 21.8% MVIC respectively. This low EMG activity was comparable to those of Ayotte et al. (2007) although their exercises were performed on a single leg. During such an exercise, the BF activity was also deemed low by Caterisano et al. (2002) and Kvist and Gillquist (2001).

There were no further differences between the four squat-based exercises. Therefore, it can be stated that additional resistance equalling 40% body mass on a barbell did not make any additional increases to BF activity. Only the back squat position was tested so the results cannot be generalised to other types of squat. This differed to the findings of Rao et al. (2009) who concluded that where possible, it was advantageous to add external resistance to rehabilitation exercises to heighten the neuromuscular activity, increasing the stability of the joint. Shields et al. (2005) also highlighted that external resistance was able to increase the hamstring activity during a single-leg squat. When comparing to the literature, it was evident there were methodological differences between the amount of weight added, the depth of the squat and the form of resistance, whether this be a rucksack (Baffa et al., 2012; Ninos et al., 1997) or barbell (Hamlyn et al., 2007; Wright et al., 1999). Caution should, therefore, be taken when comparing previous literature to the current result of 35.6% MVIC for the BF during the weighted squat. Having said this, the value of 35.0% MVIC observed by Andersen et al. (2006) was remarkably similar. Their result was for a squat to 100° of knee flexion, rather than our 90°, and subjects lifted weight equal to their 10 RM as opposed to our 40% body mass. When previous literature investigated different depths of squats using a single load throughout all the conditions, like Caterisano et al. (2002), discrepancies may become evident because the resistance was not relative. For example, the deeper the squat, the harder it is to lift the same load, so in future, it is important that the dependant variable accounts for just the depth of squat and not the load (Clark et al., 2012).

In the current study, both the mini squat and full squat produced significantly less mean EMG activity for the BF than the raised bridge and single-leg bridge. The BF is a biarticular muscle, working to extend the hip and flex the knee (Schoenfeld, 2010) thus activating differently during the two types of exercises depending upon the hip, trunk and knee positions. During squat exercises, the role of the BF is to act as the agonist for hip extension and the antagonist for knee flexion (Wretenberg et al., 1996; Wright et al., 1999). Because of the vertical stance, the quadriceps work eccentrically to lower the body and flex the knees so the main role of the BF at the knee is to counteract the anterior tibiofemoral shearing forces (Stuart et al., 1996). However, the bridge exercise incorporates hip extension against gravity, and minimal quadriceps activity (Andersen et al., 2006; Ekstrom et al., 2007), meaning the hamstrings are the primary mover for the knee joint too. It can therefore be recommended to complete bridging exercises as opposed to squats when attempting to activate the hamstrings. Ryu et al. (2011) investigated the traditional supine bridge versus a bridge with feet up against the wall and concluded the first method was more effective at activating the BF, showing the body position and gravitational forces had an impact on the mean EMG activity of this hamstring. Foot position was also a confounding factor as it was determined that rotating the foot affected the medial to lateral hamstring ratio (Lynn & Costigan, 2009). During the present study, foot position remained neutral during all exercises.

The mean EMG activity for the BF during the bridge exercise was 30.5% MVC, thus fairly similar to that of Ekstrom et al. (2007) who identified a value of 24.0% MVIC. Although not significantly different for the BF, the single-leg bridge produced a greater mean EMG activity of 42.3% MVIC. This was marginally higher but again similar to the value of 34% MVIC found by Andersen et al. (2006). This slight difference could have been due to their MVIC for the BF incorporating the mean of a maximal contraction at both 10° and 90° of knee flexion, as opposed to our 45° MVIC value. Likewise to the GMax, the position of the pelvis may have affected the hamstrings as Ishida et al. (2011) identified both an anterior and posterior tilt increased the medial hamstring activity. Their study only recorded data for one repetition, though, so the reliability of the study was undetermined.

No differences were evident between the mean EMG activity during the 'late' stage exercises for the BF so these exercises may be suitable for use within the same phase of the rehabilitation process. The step up and lateral step up activated the BF to the same level, which reflected previous results by Ayotte et al. (2007). However, Simenz et al. (2012) concluded a step up was more efficient at recruiting the BF compared with a diagonal step up, due to the sagittal plane motion providing the hamstrings with an advantage for their line of action. The results in the current study for the step up (21.4% MVIC) and lateral step up (23.7% MVIC) were slightly greater than previously reported by Ayotte et al. (2007) and Ekstrom et al. (2007). The step height of 30 cm used in the present study were based on the findings of Ekstrom et al. (2012) but the previously mentioned studies with lower EMG activity used a smaller step height. It can therefore be hypothesised that using a greater step height may be beneficial during rehabilitation for activating the biceps femoris. Finally, Ekstrom et al. (2007) placed the hamstring electrode in the middle of the muscle for the hamstrings as a group, rather than focussing on the BF, which may have affected results.

The current study identified a mean EMG activity of 40.8% MVIC for the BF during a deadlift, which was greater than the result by Escamilla et al. (2002) of 28% MVIC. However, there was a difference between the training status of the subjects with Escamilla using Division 1-A collegiate American footballers, whereas the present study used recreational athletes with a minimum of 1 hour of sport each week. The training status of the subjects was shown to affect muscle activity and recruitment (Brandon et al., 2011). The weighted squat and deadlift produced significantly less mean EMG activity for the BF than the counter movement jump. During the counter movement jump, triple extension occurred at the hip, knee and ankle, so the BF activated as the agonist to extend the hip prior to take-off and then controlled hip flexion after landing. This muscle also worked as the antagonist to control the movement at the knee. Trunk flexion was only controlled through verbal feedback but there were no objective measurements during the present study. This may have altered the results because it was identified by Kulas et al. (2010) that when the trunk was loaded, those participants who responded by flexing the trunk had increased hamstrings EMG activity to those who did not overly flex the trunk. Due to the hamstrings' attachment onto the ischial tuberosity, this muscle group is recruited to control trunk flexion (Lieberman et al., 2006). Farrokhi et al. (2008) also recognised the effect of a forward lean during lunges to increase BF activity. When prescribing rehabilitation exercises for the BF, it is therefore crucial to consider the trunk position and although out of the realms of the current study, trunk flexion may help to increase the BF activity. The high value of 59.2% MVIC for the BF during the counter movement jump was in accordance to previous research which had found that training programmes with a large focus on 'jumps for height' increased hamstring strength after a seven week intervention (Myer, Ford et al., 2006).

6.1.4 Rectus Femoris

The straight leg raise produced significantly greater mean EMG activity for the RF than the mini squat. As the RF is a biarticular muscle, flexing the hip and extending the knee (Hagio et al., 2012), the straight leg raise incorporated both actions against gravity and was therefore more specific to this muscle than the mini squat. The quadriceps setting produced less mean EMG activity than the straight leg raise for the RF but this was not a significant difference. The result of 22.4% MVIC found in the current study for the quadriceps setting was similar to that of Andersen et al. (2006), who identified the value of 24.0% MVIC during the same exercise. From these results, it can be recommended to perform the straight leg raise as a non weight bearing 'early' stage exercise for the RF. The quadriceps setting exercise may be beneficial to test isometric control of the overall quadriceps prior to performing other exercises, but may not be effective for RF strengthening purposes.

The bridging exercises produced significantly less mean EMG activity than all of the other exercises for the RF, VMO and VL. This was to be expected because the gravitational forces meant the hip flexion and slight knee extension components were completed eccentrically by the hip extensors and knee flexors, namely the GMax and hamstrings, rather than actively by the quadriceps (Ryu et al., 2011). Therefore, it is not suggested to complete these exercises when needing to strengthen the RF.

For the 'intermediate' stage exercises, the mini squat produced significantly less mean EMG activity than the full squat and lunge (14.9% MVIC, 36.5% MVIC and 39.2% MVIC respectively). During the full squat and the lunge, there were greater knee flexion angles so the quadriceps had to control the centre of mass to a lower depth (Bolgla et al., 2008). Jakobsen et al. (2013) also found the RF muscle activity increased with larger knee flexion angles, especially at the point in the exercise where the movement was reversed. The current study did not specifically look at the muscular activity during different depths but overall, the results concurred. It can therefore be proposed that where possible, the knee should be flexed to a larger range when trying to activate the RF to a greater extent.

There were no significant differences between the mean EMG activity for the step up, lateral step up, step down and single-leg squat so it can be suggested that these exercises can be used at a similar stage of rehabilitation for the RF. For the lateral step up, this study identified a mean EMG activity of 37.7% MVIC, which appeared lower than values previously described by Ekstrom et al. (2012), with their results being 58% MVIC when using a 30 cm step and 46% MVIC when using a 20 cm step. This difference may have been due to Ekstrom et al. (2012) only analysing a 0.25 second time frame surrounding the peak activity, whereas the current study incorporated data for the full 2 seconds. The single-leg squat mean EMG activity for the RF of 31.9% MVIC was more similar to the reported values by Boudreau et al. (2009) (26.7% MVIC) and Dwyer et al. (2010) (30.8% MVIC for females). The gender differences observed by Dwyer et al. (2010) will be discussed in Section 6.3 (page 98).

When focussing on the 'performance enhancing' exercises, the deadlift produced significantly less mean EMG activity than the weighted squat for the RF. In the study by Escamilla et al. (2002), the RF produced less mean EMG activity than the vasti muscles during the deadlift, which may have been due to the biarticular function of the RF. The authors of their study attributed this to the fact that if there was an increased hip flexor torque, the associated hip extensor torque would also have had to increase. The explosive jumps (single-leg vertical jump and counter movement jump) produced significantly greater mean EMG activity than both the weighted squat and deadlift. During the jumps, there was a stretch-shortening cycle of the quadriceps which enabled an increase in the muscle force from the eccentric contraction followed immediately by the concentric contraction, rather than just a concentric contraction alone (Enoka, 2008). The single-leg vertical jump and counter movement jump can therefore be recommended as suitable exercises to strengthen the RF during the latter stages of rehabilitation.

6.1.5 Vastus Medialis Oblique

There was a significant difference in the mean EMG activity between the two non weight bearing exercises. The straight leg raise produced less mean EMG activity than the quadriceps setting exercise. During rehabilitation, it is usually assumed that the patient should be able to perform an isometric contraction of the quadriceps prior to raising the leg off the ground as seen in a straight leg raise. However, the current results showed that for the VMO, less activity of the muscle occurred during the straight leg raise. These results were similar to that of Kushion et al. (2012), who concluded that short arc quadriceps extension was more efficient for training both the VMO and VL than a straight leg raise. However, the results differed from those of Soderberg et al. (1983). Although Bolgla et al. (2008) only measured the EMG activity during the straight leg raise, and not the quadriceps setting exercise, their results for the VMO were considerably greater (26.0% MVIC) than those in the current study (9.0% MVIC). All straight leg raises were completed with an extended knee but the difference in mean EMG activity may have been due to less emphasis on the knee extension and more on the hip flexion, activating the RF, in the current study, whereas Bolgla et al. (2008) extended the knee isometrically for one second, prior to raising the leg.

The mini squat mean EMG activity for the VMO was significantly less than for the full squat and lunge. This was likely to have been affected by the range of motion at the knee, with both the full squat and lunge incorporating 90° of knee extension, whereas there was only 45° range during the mini squat. This meant the full squat and lunge had an increased applied torque due to the gravity (Bolgla et al., 2008). These conclusions were similar to that of Jakobsen et al. (2013) who identified the more flexed the knee became, the more the VMO and VL EMG activity increased. Due to the lower amount of musculature activity for the VMO, the mini squat (29.3% MVIC) may be a beneficial exercise to complete as an introduction to partial weight bearing during the rehabilitation process, followed by the full squat and lunge. There were no differences between the mean EMG activity for the full squat and lunge so it can be suggested that these activate the VMO to a similar extent so can be used simultaneously during rehabilitation. Only the front leg of the lunge was tested though, so this suggestion may be altered if both legs were measured.

No differences were evident between the mean EMG activity for the VMO during the 'late' stage exercises so it can be recommended that they would have similar effects on the VMO during rehabilitation. The exercises all involved flexion and extension of the knee on a single-leg stance. Although slightly different EMG activities were present, the results in the current study followed a similar trend to those of Bolgla et al. (2008), by which these single-leg exercises produced greater mean EMG activity for the VMO than double leg squats, for example. The mean EMG activity for the lateral step up in the current study was comparable to that of Ekstrom et al. (2007) (94% MVIC and 85% MVIC, respectively). From the results, adding in a lateral component during the lateral step up compared with the front step up, made no difference to the VMO activity so either exercise can be used during rehabilitation dependent upon which appears to be more sports specific for the individual. The step down exercise incorporated a higher requirement for eccentric control of the quadriceps so although the EMG activity appeared similar to the step up exercise, during progressive rehabilitation, it may be advised to complete the step up first. In the study by Bolgla et al. (2008), the EMG activity was considerably lower with the step down recording values of 27% MVIC and the step up producing values of 32% MVIC. The differences in the level of results may have been due to the height of the step; the lower the step, the greater the likelihood the tibia was displaced anteriorly over the foot, thus increasing the weight transfer and potentially the EMG values of the quadriceps (Mathiyakom et al., 2005).

The mini squat produced significantly less mean EMG activity than the full squat, weighted squat and single-leg squat for the VMO. There were no significant differences between the EMG activity for the other three types of squats. It can therefore be speculated that during a squat to 90° of knee flexion, a barbell containing an additional 40% body mass, cannot be used to increase the VMO activation. For the VMO, completing a double-leg squat to 90° was similar to a single-leg squat to 60° so either are suggested during the rehabilitation of this muscle.

For the 'performance enhancing' exercises, the deadlift produced significantly less mean EMG activity than for the counter movement jump and single-leg vertical jump. Toumi et al. (2007) compared a double-leg drop jump with a single-leg landing and concluded that the VMO was activated to a greater extent during the single-leg movement. They proposed that in a dynamic situation, it may be more challenging to control the patella when on one leg as this position may destabilise the knee to a greater extent. Although not significantly different in the current study, the single-leg vertical jump did produce a greater EMG activation of the VMO compared to the counter movement jump.

6.1.6 Vastus Lateralis

Similarly to the VMO, the straight leg raise produced significantly less mean EMG activity than the quadriceps setting, even though both exercises included knee extension in a non weight bearing position. Although the VMO and VL would have been activated to extend the knee, this was not the main focus of the straight leg raise, hence the RF and although not tested, it can be hypothesised the iliopsoas, were the primary movers. The mean EMG activity of the VL during the quadriceps setting exercise was comparable to the results by Andersen et al. (2006) (30% MVIC and 32% MVIC, respectively). These results indicated that the quadriceps setting exercise was below the 40% MVIC threshold to bring about a muscular adaptation in strength (Andersen et al., 2006; Escamilla et al., 2010). However, it may be a useful exercise to begin rehabilitation as it is in a non weight bearing position.

The VL reacted in a similar way to the VMO during the 'intermediate' stage exercises; the mini squat, once again, generated less EMG activity than the full squat and lunge. The latter two exercises incorporated a larger range of motion at the knee which would have elongated the quadriceps, thus affecting the muscle's length-tension relationship (Bolgla et al., 2008). The VL was also investigated during a lunge exercise in a study by Farrokhi et al. (2008). The mean EMG activity was greater in the current study (69% MVIC versus 45.6% MVIC). Although both studies used relative step lengths for each subject, the data analysis procedures differed for the time frame which was used, potentially affecting the results. Also, in the current study, the subject started and finished with their feet apart, to control the step distance and ensure it was a smooth motion, as identified during pilot testing. This could have meant the EMG activity was greater for the VL due to only measuring the middle section of the exercise. From these results, though, the full squat and lunge should be used as a progression to the mini squat during rehabilitation.

For the two step ups and single-leg squat, no differences were identified for the VL mean EMG activity. It can therefore be proposed that these exercises would have a similar training effect on the VL so could be used interchangeably within the rehabilitation process. Both step ups, the step down and single-leg squat all involved controlling the centre of mass over a small base of support (Bolgla et al., 2008). This is an important concept to incorporate into rehabilitation as walking, running and stair climbing all involve a single-leg stance and the weight transfer between the two lower limbs.

Once more, the deadlift produced significantly less VL mean EMG activity than the counter movement jump and single-leg vertical jump. With the deadlift, the trunk position meant there was a higher focus on the hamstrings working eccentrically to lower the bar (Wright et al., 1999), hence there was less emphasis on the quadriceps. Whereas with the two types of jump exercises, both the VMO and VL had to powerfully contract to extend the knee just before take-off, then contract eccentrically to control the landing. There were no significant differences between the mean EMG activity for the counter movement jump (91.4% MVIC) and single-leg vertical jump (100.9% MVIC) for the VL. It can therefore be suggested that performing either one of these exercises may be equally as beneficial to strengthen the VL. The extremely high EMG activity during the jumps shows these exercises should only be performed during the late stage of rehabilitation or as part of a strength and conditioning programme to enhance performance.

6.2 Muscle Ratios

6.2.1 Vastus Medialis Oblique:Vastus Lateralis Ratio

Previous literature has stated that the VMO specifically activates during terminal knee extension due to less lateral restraint from the femoral groove during this range, hence there is more emphasis on this muscle acting as a medial patella stabiliser (McConnell, 1996; Toumi et al., 2007). Also, it was proposed that the VMO function became more significant during this final 15˚ of knee extension range due to a 60% increase in force production from the quadriceps (Sakai et al., 2000). Interestingly, the results from the current study opposed this because the quadriceps setting and mini squat, which both focussed on the final phase of knee extension were two of only four exercises to produce VMO:VL ratios below the equal threshold of one. This highlighted that the VMO activated to a lesser extent than the VL so it cannot be recommended to use this exercise when it is necessary to specifically activate the VMO.

Similarly to the results from this study, Balogun et al. (2010) and Mirzabeigi et al. (1999) also identified that the quadriceps setting exercise activated the VL to a greater extent than the VMO, indicating a VMO:VL ratio of less than one. The ratio for this exercise within the current study was significantly lower than the results for eight of the other exercises, including the mini squat, full squat, single-leg squat and step-based exercises. Although this result opposed the aforementioned literature, it was actually in agreement with other studies, which concluded the VMO activated to the greatest extent during mid-range, namely 60° (Baffa et al., 2012; Shenoy et al., 2011; Tang et al., 2001). Shenoy et al. (2011) specifically identified the VMO had greater integrated EMG at 60°, followed by at 90°and lastly at 30°, which reflected the knee extension peak torque. More functional testing was completed by Tang et al. (2001) and results showed that a squat-to-stand exercise between 0-60° produced a greater VMO:VL ratio than a 0-15° squat-to-stand exercise. The single-leg squat to 60° of knee flexion produced the highest VMO:VL ratio in the current study. When considering the range of motion at the knee during rehabilitation exercises for patients with PFPS, the amount of patellofemoral joint stresses need to be addressed. Steinkamp et al. (1993) advised CKC exercises between 0-45° to reduce the risk of pain from increased forces. With this in mind, the results from the current study and those of Tang et al. (2001) and Shenoy et al. (2011) cannot state that all exercises to increase the VMO strength should be to 60° of knee flexion, but when pain-free, this may be a beneficial range to progress to.

From a clinical viewpoint, the quadriceps setting movement represents the exercise given by many clinicians to target the VMO (Lieb & Perry, 1968; Witvrouw et al., 2000). The manual muscle test used for the VMO is performed in a similar manner, too (Hertling & Kessler, 2006). There has also been an intervention study, which incorporated short arc quadriceps exercises, leading to a decrease in pain scores after six weeks. However, the results were not definitive in saying that the exercises specifically strengthened the VMO (Minoonejad et al., 2012). The current trends in sports therapy may therefore need to be reassessed in the future, in light of these new findings.

Bennell et al. (2010) completed a study in which OKC exercises between 0-10° of knee flexion were compared to more functional 0-90° of knee flexion exercises. The second set of exercises increased the timing of the VMO in relation to the VL, whereas the short arc OKC exercises strengthened the overall quadriceps. These results were compared favourably to the current study; the more functional exercises, which utilised a larger range of knee motion, were most effective for activating the VMO. In view of the VMO:VL results, the highest ranked exercises were all of this nature. It must be stated that the current study did not measure the timing of the two muscles during the rehabilitation exercises, unlike Bennell et al. (2010), but this could be an area for future research.

The counter movement jump produced the VMO:VL ratio closest to the value of one, indicating that in healthy participants, this type of exercise had the optimal balance between the two vasti muscles. In such an exercise, there was a high reliance on the co-activation of the lower limb muscles in general, shown by the current results for the HS:Quads ratio being 0.99 as well as the VMO:VL ratio being 0.97. This exercise may be beneficial during the latest stages of rehabilitation, or during conditioning sessions as a preventative measure (Waryasz & McDermott, 2008). The fact that the VMO:VL ratio was equal in healthy participants should be reassuring for clinicians so this type of exercise should not pre-dispose to an imbalance within the vasti group, leading to PFPS. When comparing the VMO:VL ratios for the counter movement jump and single-leg vertical jump, the single-leg version was greater (0.97 and 1.08 respectively) but this was not a significant difference. As previously mentioned, greater ratios were potentially recorded for the VMO and VL during a single-leg stance due to the increased need for patella stability when the centre of mass was displaced laterally (Toumi et al., 2007). This type of exercise could therefore be used during the latter stages of rehabilitation with an athlete who had suffered from a weakened VMO.

In healthy individuals, it was previously reported that step up and step down tasks adopted a VMO:VL ratio greater than one (Crossley et al., 2002). This was similar to the VMO:VL ratios within the current study (step up: 1.21, step down: 1.19). Variances between the ratios differed slightly, but this could have been due to differences such as the step height, the EMG data analysis process and EMG electrode properties. If an athlete suffers from PFPS, stair climbing has been identified to aggravate symptoms (Callaghan et al., 2009), so this type of exercise should either be incorporated into an exercise programme once pain free or used more as a preventative exercise. To reiterate further, it is important that the athlete does not experience pain during rehabilitation because it was identified that pain inhibition and swelling can affect the VMO activation (Hopkins et al., 2001).

6.2.2 Hamstrings:Quadriceps Ratio

The HS:Quads ratio is an important factor for patients with knee osteoarthritis and during prevention strategies and rehabilitation of those with an ACL sprain (Hortobágyi et al., 2005; Myer et al., 2009; Urabe et al., 2005). The role of the ACL is to prevent anterior displacement of the tibia on the femur (Herman et al., 2008), thus, the hamstrings must contract to counterbalance the anterior forces produced by the quadriceps (Stuart et al., 1996). When considering the HS:Quads ratio, a value greater than one indicates the BF was activated to a greater extent than the mean ratio for the RF, VMO and VL. A ratio of less than one highlights the quadriceps were activated to a greater extent than the BF.

The bridge, single-leg bridge and raised bridge were hamstring dominant CKC exercises, with the HS:Quads ratio varying between 10.2-16.6 for the three exercises. These were significantly greater HS:Quads ratios than all of the other exercises ($p < 0.05$). In the past, there have been varying views between whether ACL rehabilitation exercises should be OKC or CKC, due to the need to restrict the amount of anterior tibial translation (Beynnon et al., 1997; Kvist & Gillquist, 2001; Lysholm & Messner, 1995). It was proposed that during loaded CKC exercises, the tibiofemoral joint had a higher compressive force, thus allowing less tibial translation (Kvist & Gillquist, 2001; More et al., 1993). With this in mind, the bridges may therefore be an effective exercise to preferentially strengthen the hamstrings, post ACL injury. As previously discussed, due to the supine positioning during the bridges, the hip and knee flexion/extension were mainly performed by the posterior muscles including the hamstrings and GMax to counteract the gravitational forces. Within the bridging exercises, the raised bridge produced a significantly higher HS:Quads ratio than the single-leg bridge ($p = 0.038$). The raised bridge incorporates a hip extension torque from a greater hip flexion angle, which may have been the reasoning behind the hamstrings being recruited to a greater extent; the proximal end of the hamstrings were stabilised, allowing the distal end of the muscle group to work more efficiently (Oliver & Dougherty, 2009).

The four squatting exercises produced HS:Quads ratios between 0.58-0.88, indicating low BF activity with respect to the quadriceps. During the decent phase of the squat exercises, knee flexion is eccentrically controlled by the quadriceps to lower the body's centre of mass (Bolgla et al., 2008). The hamstrings work as the antagonist to control the knee flexion angle (Wright et al., 1999) and reduce anterior translation of the tibia (Stuart et al., 1996). This may explain the low HS:Quads ratios observed during the squats in this study and previous literature (Ayotte et al., 2007; Caterisano et al., 2002). However, the results may have been different if a wall squat was completed, which involves the centre of mass being posterior to the feet (Kvist & Gillquist, 2001).
Both the deadlift and weighted squat involved lifting a barbell equivalent to 40% of the subject's body mass. However, the deadlift produced a significantly greater

HS:Quads ratio than the weighted squat ($p = 0.001$). During the deadlift, the hamstrings acted as the synergist to the GMax to help control the trunk flexion, and extend the hip during the ascent (Lieberman et al., 2006). The quadriceps were the antagonist during the deadlift, whereas with the weighted squat, as described in the previous paragraph, they worked eccentrically to control the knee flexion. Therefore, it can be proposed that the trunk position and greater emphasis on the 'upward' hip extension phase during the deadlift were the reasons behind the differences in the HS:Quads ratios (Hales et al., 2009). A very similar HS:Quads ratio was identified by Ebben et al. (2009) of roughly 1.5, compared to the value of 1.2 in the current study. Although a difference was evident between these values and those of Begalle et al. (2012), their overall observation was comparable, with the deadlift producing a greater ratio than that produced by the squat and lunge. The differences in actual values may have been as a result of their study using the average of the vastus medialis, VL, BF and medial hamstrings within the ratio calculation as opposed to our ratio which comprised of the VMO, VL, RF and BF. Finally, Wright et al. (1999) performed a similar study and concluded the stiff leg deadlift recruited less quadriceps and more hamstring activity, which concurred with the current study. From these results, the deadlift may be an effective exercise during the final stages of ACL rehabilitation (Escamilla et al., 2002), as opposed to squats.

Similarly to the VMO:VL ratio, the counter movement jump generated the HS:Quads ratio closest to the value of one (0.99). Chappell et al. (2007) concluded that during a vertical stop jump, athletes should train the neuromuscular patterns throughout the preparation for landing as females recruited lower hamstring activity at this stage compared to the quadriceps, leaving them at risk of injuring the ACL. From the current results, the muscle onset timings for the hamstrings and quadriceps cannot be stated but when investigating the overall magnitude throughout the whole exercise, the counter movement jump appeared to be the most efficient exercise to co-contract the hamstrings and quadriceps to the same intensity in healthy participants. Within the seven week training programme by Myer, Ford et al. (2006), the increase in hamstring strength could have been partially attributed to the large focus on this type of jump.

During the latter stage of rehabilitation, single-leg activities are important due to replicating functional movements but also to allow the clinician to identify any side-to-side asymmetries, which may include an overreliance on the unaffected limb (Begalle et al., 2012; Myer, Paterno et al., 2006). The single-leg vertical jump and single-leg squat produced ratios of 0.86 and 0.88 respectively, indicating these exercises had a higher focus on the quadriceps and less on the hamstrings compared to the counter movement jump. These results differed from those of Kean et al. (2006) who concluded the co-activation ratio between the hamstrings and quadriceps was approximately 36% higher during the single-leg jumps versus the two footed jumps. Their study had a greater emphasis on the analysis of the landing, though, rather than the full jump, which may have affected the results. Furthermore, their MVIC was completed in a standing position with the pelvis unsupported, which differed from the method in the current study. Shields et al. (2005) highlighted that additional external resistance was able to increase hamstring activity during a single-leg squat, so this may be the preferential way for an athlete to perform single-leg squats when recovering from an ACL injury as opposed to the type which was tested in the current study. It is essential that an athlete has correct biomechanics during such movements before returning to play, but due to the HS:Quads ratios found in the present study, it can be proposed that extensive hamstring strengthening through bridges, deadlifts and counter movement jumps should be undertaken prior to participating in these single-leg exercises. An example rehabilitation programme by Myer et al. (2008) for post-ACL reconstruction surgery included all of the exercises aforementioned, although the single-leg isometric squat was completed earlier in the process than thought from this study's results.

6.2.3 Gluteus Maximus:Biceps Femoris Ratio

Jonkers et al. (2003) identified that strength impairments of the dominant muscles led to an increased activation of the synergists. The GMax and hamstrings work together to extend the hip but a decrease in GMax strength may result in the athlete becoming hamstring dominant (Chance-Larsen et al., 2010; Lewis et al., 2007). The hamstrings cannot always withstand the extra load, thus potentially causing an injury (Devlin, 2000). Recurrence rates of hamstring strains are alarmingly high (Petersen & Hölmich, 2005). It is therefore important to re-educate the GMax and strengthen this muscle to increase pelvic stability and the transfer of forces between the lower and upper limbs (Kibler et al., 2006) as well as to prevent lower limb injuries (Reiman et al., 2009).

The step up produced the highest GMax:BF ratio of 1.55. The lateral step up and step down generated similarly high ratios of 1.50 and 1.42 respectively. The ratios for these three exercises were significantly higher than the GMax:BF ratios for the raised bridge, mini squat, full squat and weighted squat. Furthermore, the ratios for the step up and lateral step up were also higher than for the deadlift. The step-based exercises incorporated a single-leg stance, thus activating the GMax to resist internal rotation of the femur by providing an external rotation torque. In terms of the hip extension motion, the hamstrings acted as the synergist to the GMax. The step up involved hip extension from roughly 90° of hip flexion to a neutral alignment. The centre of mass was displaced anteriorly, with respect to the weight bearing foot, during the upward phase, which may have increased the hip thrust action. This appeared to increase the amount of GMax activity in comparison to the BF. In a functional study, it was identified that during stair ascent, the lower section of the GMax acted as the primary hip extensor (Lyons et al., 1983). This finding compared favourably to the present study in that raising the body up onto a step recruited the GMax as the prime muscle to extend the hip, rather than the hamstrings. Therefore, when clinicians require the athlete to activate the GMax with less hamstrings involvement, step up exercises are recommended, based on these findings.

During the prone hip extension, a value of 1.0 was identified for the GMax:BF ratio. As previously stated, the majority of the past literature observed the onset timing of the muscles rather than the magnitudes during this movement (Bruno et al., 2008; Chance-Larsen et al., 2010; Lehman et al., 2004). However, a recent study investigated both types of measurements during prone hip extension but when combined with knee flexion, rather than knee extension (Kang et al., 2013). The effects of an abducted hip position were examined, compared to a neutral hip position. The authors concluded that when the hip was abducted, the GMax:BF ratio increased from 1.13 (neutral position) to 2.06 (30° of hip abduction). Direct comparisons cannot be made between the results in the current study and the neutral position in their study due to their movement having a decreased lever length, affecting the torque. Kang et al. (2013) proposed the increased findings during the abducted position were due to the oblique fibre arrangement of the GMax matching the line of action of the muscle (McAndrew et al., 2006). Further research should be completed on this area but from the current results, the overall action of a prone hip extension movement activated the GMax and BF to similar amounts in healthy participants. It can therefore be used to activate both muscles, without favouring one hip extensor over the other.

When considering the GMax:BF ratio, the raised bridge produced significantly lower results than the single-leg bridge, meaning there was relatively less GMax activity and greater hamstring involvement during the raised bridge. When considering the hamstrings, by placing the hip in a more flexed position, as seen during the raised bridge, the proximal aspect of the muscle was stretched, allowing greater muscular advantage at the knee (Oliver & Dougherty, 2009). When considering the bridging exercises, it can therefore be suggested that when there is a greater focus on the activation of the GMax, the single-leg bridge should be performed during rehabilitation, whereas when requiring an increase in hamstrings activity, the raised bridge should be prescribed.

6.3 Gender Differences

The GMax produced significantly greater mean EMG activity for the females compared to the males during the raised bridge, lunge, step down, counter movement jump and single-leg vertical jump. Dywer et al. (2010) observed similar trends with the overall concentric GMax activation being greater during a single-leg squat, lunge and step-up-and-over in females compared to males. When focussing on the eccentric GMax EMG activation, females only exhibited greater values during the single-leg squat and lunge. The knee flexion angles differed between the sexes in their study, which was proposed by their authors to be due to differences in strength, allowing men to squat further. The variation in strength between genders does not fully explain the GMax EMG activity differences though, because all values were normalised to a MVIC. Females were identified to produce greater hip extension angles during such exercises (Dywer et al., 2010) and whilst running (Ferber et al., 2003) which could have been the reasoning for the increased GMax EMG activity due to this muscle controlling the additional range of motion (Chumanov et al., 2008). Hip angles were not measured in the current study so it cannot be determined whether this theory explains our results. Having said this, when rehabilitating athletes, the hip flexion and extension range of motion may be something to consider, ensuring males are able to perform throughout the same range as females. Increased tightness in the hip flexor musculature was evident in subjects with decreased hip extension range of motion (Schache et al., 2000). This goes outside the realms of this study, but pelvic tilt and lumbopelvic dysfunction should be analysed for both genders as this was observed to alter muscle activation of the hip and trunk muscles (Tateuchi et al., 2012).

Bouillon et al. (2012) examined the lunge and step down but no significant differences were evident between the genders for the EMG activity of the GMax, GMed, BF and RF. These results differ from those observed in the current study. Bouillon et al. (2012) used an individual step height for each subject, which corresponded to 25% of the subject's leg length. Although the current study used a standardised step height, there were no significant differences in knee flexion angles between males and females so this may not have affected the results to a great extent. It should be noted that Bouillon et al. (2012) performed the MVICs against manual resistance, which may have led to reduced effort levels depending upon the stability of the resistance and the strength of the examiner (Shenoy et al., 2011; Silvers & Dolny, 2011). The present study used a fixed resistance belt which was firmly secured and did not rely on human strength (Silvers & Dolny, 2011), thus meaning the MVICs may have been a better reflection of maximal strength.

The RF displayed differences in mean EMG activity between the genders during the step up, step down and lateral step up. The females had greater activation of this muscle than the males. This was a similar finding to that of Dywer et al. (2010) and Zeller et al. (2003). The RF activates to control movement in the sagittal plane during exercises such as the step up and step down, but it cannot be exactly determined why the gender difference occurred without more data regarding the kinematics at the hip joint. Once again, Dywer et al. (2010) attributed the increase in RF EMG activity in females to be a result of the overall lower strength in women, thus requiring the muscle to contract to a greater extent. Pelvic tilts may have further implications for this difference between the genders; the cause-effect relationship cannot be determined but Tateuchi et al. (2012) concluded an increase in hip flexor EMG activity was associated with an anterior pelvic tilt. Further research should be conducted to investigate this point further. Although not evident in the current study, females have been previously observed to have an increased RF activation during the pre-contact period when landing from a jump, which may be a contributor to increased ACL sprains during this type of motion (Zazulak et al., 2005). This highlights the need for clinicians to observe pelvic disorders and rehabilitate this prior to the athlete obtaining any additional injuries. If a subsequent injury does occur, the therapist needs to ensure the rehabilitation also treats any underlying muscular imbalances or flexibility issues, and this may be partially related to their gender.

The GMed EMG activity was significantly greater for the females compared to the males during the lunge. This was not expected due to previous research highlighting females had greater hip adduction torques due to a lack of GMed activity (Chumanov et al., 2008; Earl et al., 2007). During the lunge, albeit a wide stance, both feet were still in contact with the floor, so the hip abductors on the dominant side did not have to solely stabilise the trunk, head and arms (Bolgla et al., 2008). The bilateral stance meant there was less demand on the hip musculature for frontal plane stability (Earl et al., 2007), which could have been the reason as to why the GMed reacted in a different way to expected. Previous studies observed no gender differences when focussing on the normalised EMG activity during single-leg exercises (Bouillon et al., 2012; Dwyer et al., 2010; Zazulak et al., 2005). However, their results were inconclusive at identifying if there were any differences in the overall strength of the GMed between the genders. Even though the current study observed a greater EMG activity in females during the lunge, when viewing the mean EMG activity of 21.1% MVIC, it cannot be suggested that this is an effective exercise when needing to strengthen the GMed.

Males had significantly greater mean EMG activity for the BF during the single-leg vertical jump compared to the females, even though there were no significant differences in the mean knee flexion angles between the genders prior to take-off. As previously stated, the results of the current study incorporated the full take-off and landing stages for this exercise so the comparison to other studies which just focussed on the landing phase should be viewed with caution. Having said this, Chappell et al. (2007) presented similar findings. Their study specifically focussed on the landing of a vertical stop jump; results showed an increase in the EMG activity of the hamstrings in the male subjects after landing, compared to the female counterparts. Ebben et al. (2010) also found comparable results with the males demonstrating increased hamstrings activity after landing. However, no gender differences were evident for the magnitude of the hamstrings activation prior to the floor contact. These findings may be a confounding factor for the higher incidence rates of ACL sprains in females due to the lower hamstrings activity being less able to withstand the anterior tibial translation forces produced by the quadriceps (Stuart et al., 1996). From the current results, it is therefore crucial that females undertake a strengthening programme including bridges and deadlifts as discussed in Section 6.1.3 (page 83), prior to participating in plyometric based exercises.

6.4 Limitations

There were limitations present within the current study. The trunk position can affect the activation of the gluteals and hamstrings, thus the examiners attempted to standardise the positioning during the exercises with the use of verbal and physical cues. However, no objective measures were used to monitor the trunk flexion or extension, which may have affected the results. However, the findings more closely replicate the way in which exercises are performed in a clinical setting.

Standardised protocols were followed; knee flexion angles were standardised during the squats and bridges, the hip abduction angles were controlled during the non weight bearing exercises, and the lunge step-distance was relative to the subject's leg length. However, the height of the step used during the step up, step down and lateral step up was not normalised in this way. There were no significant differences between the knee flexion angles for males and females during the step-based exercises though, which suggests task demands were similar for subjects of differing heights. The EMG activity may therefore not have been affected. Finally, it was assumed that the subjects generated a true MVIC for each muscle. The study attempted to control this by providing standardised verbal encouragement, which has been shown to increase contraction levels. The pilot testing also investigated the most appropriate positions to obtain the greatest EMG activity. However, even with these procedures in place, the single-leg vertical jump exceeded 100% MVIC for the VMO and VL.

6.5 Clinical Implications

The single-leg vertical jump can be used as a late stage exercise to strengthen the GMax. Although this was the only exercise to recruit the muscle above the 40% MVIC threshold for strength gains, the counter movement jump and single-leg bridge may also be used to increase the endurance of the muscle to prevent lower limb injuries and stabilise the pelvis during functional activities.

The GMed was identified as being activated to the greatest extent during single-leg activities, where the role of supporting the pelvis and upper body became more significant. The dynamic single-leg vertical jump was the only exercise to produce results in excess of 40% MVIC so may be used to strengthen this muscle in order to decrease injury or improve performance. Single-leg squats, single-leg bridges and hip hitches are also recommended for clinicians to prescribe to activate this muscle for stability and postural purposes. If the athlete is unable to fully weight bear, the single-leg bridge is the most appropriate alternative.

The explosive jumps, single-leg bridge and deadlift produced the greatest EMG activity for the BF so these exercises may be used within the latter stages of rehabilitation for hamstring strains. The trunk position is vital to consider because the hamstrings control trunk flexion, thus when this muscle group is placed on a stretch at the proximal end, they have a mechanical advantage at the knee.

The VMO and VL produced similar trends within the 20 exercises during this study, whereas the RF EMG activity was more individual. It is well-known that the RF acts over the hip and knee; these results show this fact cannot be overlooked and so during rehabilitation, this muscle needs to be treated differently to the rest of the vasti group. The straight leg raise produced mean EMG activity 2-3 times greater for the RF than the VMO and VL so it is an appropriate non weight bearing exercise during early rehabilitation for the RF. The exercises which activated the RF to the greatest extent include the counter movement jump, single-leg vertical jump and weighted squat. These exercises can therefore be used during late stage rehabilitation to activate this muscle.

The VMO mean EMG activity was greatest during the single-leg vertical jump. Other exercises, including the counter movement jump, step ups, step down and single-leg squat can also be used during late stage rehabilitation to activate this muscle. The mini squat may be effective to introduce the subject to partial weight bearing activities due to a low mean EMG activity, thus not putting excess strain on the muscle. Similar trends were seen with regards to the VL. The single-leg squat to 60° of knee flexion produced the greatest VMO:VL ratio so is an appropriate exercise to preferentially activate the VMO over the VL. The straight leg raise and quadriceps setting exercises produced the lowest VMO:VL ratio so they may not be beneficial for patients with PFPS. The more functional exercises produced a greater VMO:VL ratio; however, these may not be suitable for patients with PFPS during initial rehabilitation because pain may be reproduced due to the increased patellofemoral compressive forces. The current research indicates that when pain-free, patients with PFPS should progress to exercises with moderate amounts of knee flexion and extension, rather than focussing purely on end of range knee extension.

6.6 Recommendations for Future Research

It was important to gauge the amount of muscular activity during rehabilitation exercises on the continuum from non weight bearing movements through to those with an additional external resistance, in order to identify a baseline of measurements for healthy individuals. However, future research should investigate if the EMG activity remains similar when using a cohort of subjects who are injured and in the process of rehabilitation. Injuries may respond differently so having these findings would mean rehabilitation could be more focussed and specific than ever before. The EMG activity could also be divided to correspond with 10° sections throughout the knee flexion range of motion to determine the precise effect of each angle rather than considering the exercise as a whole.

It would also be interesting to determine strength gains, injury prevalence results and sports performance after undertaking an intervention programme involving the greatest EMG activity exercises from this study to view results in a more functional context. Finally, research could be executed to establish which muscles activate to a lesser or greater extent during these exercises if variations in technique and limb placement are evident. This would help clinicians to fully understand the process of performing exercises for prehabilitation, rehabilitation and performance enhancing purposes.

6.7 Conclusion

This study has identified the most effective exercises which can be used to strengthen the GMax, GMed, BF, RF, VMO and VL during the latter stages of rehabilitation and to improve performance. These findings were inferred by highlighting the exercises which produced the greatest mean EMG activity. For each of the muscles, an overall continuum of exercises has also been observed, which ranks the exercises based upon the normalised mean EMG activity. This will provide support for clinicians by suggesting an appropriate and progressive order for rehabilitation, prehabilitation and performance enhancing purposes. The VMO:VL ratio, HS:Quads ratio and GMax:BF ratio were also considered for specific rehabilitation.

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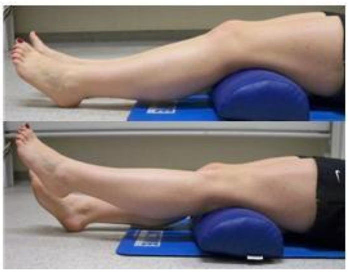
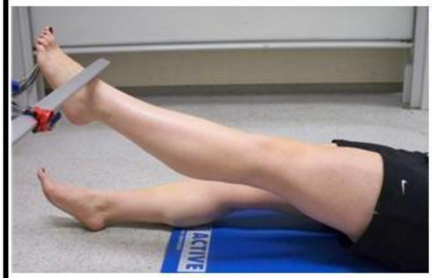
Appendices

Appendix One: Photographs of the Twenty Rehabilitation and Performance Enhancing Exercises

Early Stage of Rehabilitation Exercises



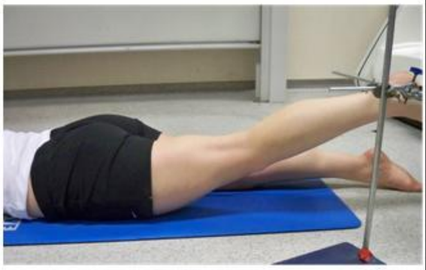
1. Quadriceps Setting



2. Straight Leg Raise

3. Side-lying Hip Abduction

4. Standing Hip Abduction



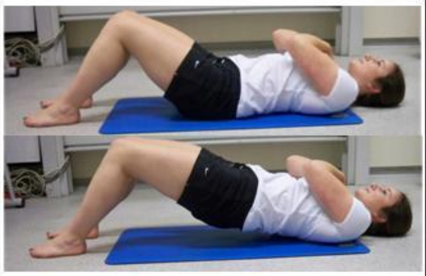
5. Prone Hip Extension

Intermediate Stage of Rehabilitation Exercises

6. Mini Squat

7. Full Squat

8. Lunge



9. Bridge

10. Single-leg Bridge

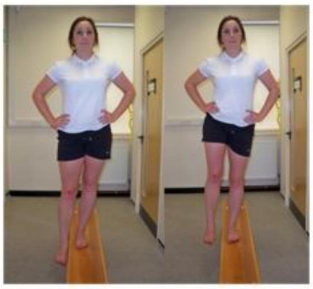
11. Raised Bridge

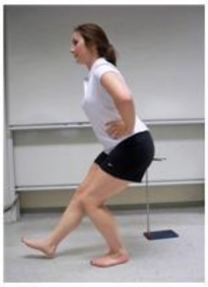
Late Stage of Rehabilitation Exercises


12. Step Up


13. Step Down


14. Lateral Step Up


15. Hip Hitch


16. Single-leg Squat

Performance Enhancing Exercises


17. Counter Movement Jump


18. Single-leg Vertical Jump


19. Weighted Squat


20. Deadlift

Appendix Two: Photographs of the MVIC Positions

Gluteus Maximus

Lying prone, knee flexed to 90°, isometric hip extension.

Gluteus Medius

Side-lying, in 20° of hip abduction, isometric hip abduction.

Biceps Femoris

Lying prone, knee flexed to 45°, isometric knee flexion.

Rectus Femoris, Vastus Medialis Oblique and Vastus Lateralis

Seated, knee flexed to 60°, isometric knee extension.

Appendix Three: Subject Briefing Form

UNIVERSITY OF HERTFORDSHIRE
FACULTY OF HEALTH & HUMAN SCIENCES
School of Life Sciences

BRIEFING DOCUMENT

TITLE OF STUDY

Electromyographic Analysis of Hip and Knee Exercises: A Continuum from Early Rehabilitation to Enhancing Performance.

INTRODUCTION

There is currently a lack of evidence for the progression of exercises used to target specific hip and knee muscles during the rehabilitation of an injury. This study is therefore going to measure the muscular activity of the quadriceps, hamstrings and gluteal muscles during twenty exercises which are currently used by medical practioners. These exercises will incorporate a range from early through to late stages of rehabilitation, and also exercises used to enhance performance. The results will endeavour to make the rehabilitation for an array of lower limb injuries more efficient in the future.

AM I ELIGIBLE TO TAKE PART IN THE STUDY?

You will be required to adhere to the following criteria. Subjects will be:

- Volunteers, recruited from personal contacts and posters around The University of Hertfordshire.
- Students at The University of Hertfordshire.
- Between 18 and 26 years old.
- A participant in over an hour of exercise each week.
- No history of surgery to the lower limb.
- No history of a knee injury in the past 12 months.
- Not suffering from a musculoskeletal injury to the lower limb within the past 6 months.
- Not suffering from any central or peripheral neurological conditions.
- No allergy to adhesive tape.
- Not currently pregnant.

Up to one hundred subjects will be recruited, both males and females, all tested individually.

WHAT IS INVOLVED?

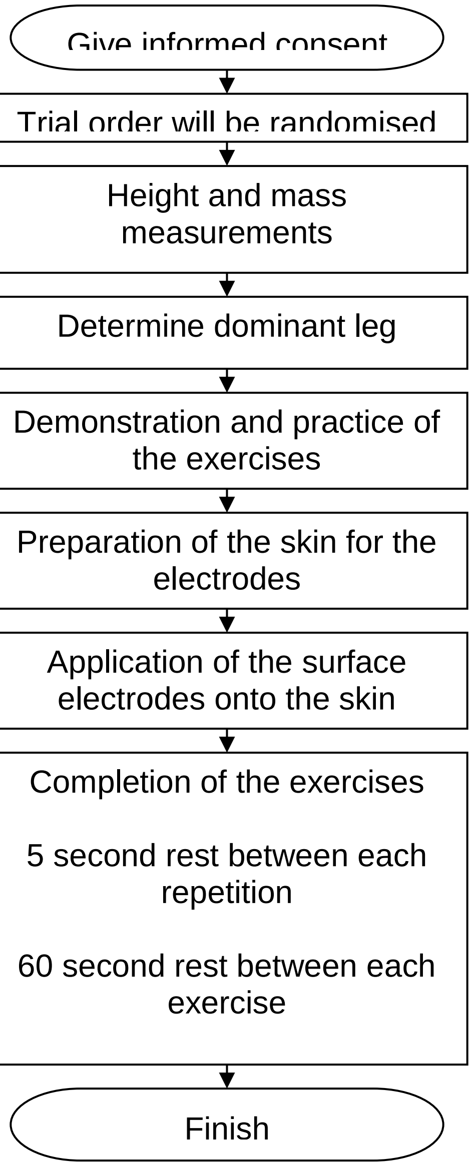
To begin with, you will read and discuss with the examiner the purpose of the testing, as highlighted in the introduction of this document. You will have the opportunity to ask questions and then once you are happy to proceed, you will sign an LEC2 consent form. Prior to the testing session you should be prepared to shave the designated areas on your upper thigh as stated by the tester. Your height and mass will be recorded and your dominant leg will be determined. You will randomly be assigned to complete the exercises in a specific order by the tester choosing numbered envelopes. You will be required to attend three testing sessions, each lasting a maximum of one hour. There will need to be a day of rest between each session.

At each session, you will complete 7 exercises; for example, a straight leg raise or a mini squat. The tester will demonstrate the exercises and then you will practice the movements twice through each. There will be methods of control in place to ensure the results are reproducible, such as when you squat, there will be a bar to show you how low to go.

An alcohol wipe will be used to clean the areas of skin on your thigh and then the electrodes will be applied. During the testing each exercise will be performed five times. The speed at which you perform these will be determined by a metronome. There will be a 60 second rest period between the different exercises.

You will be verbally debriefed after the testing has been completed to highlight the purpose of the study and to describe what the results can show. The risks to you within this study are minimal. There is a small risk of you suffering a muscle strain from completing the exercises but these risks are limited by having a 60 second rest period in between each exercise and by completing the practice exercises to ensure you have the correct technique. The surrounding area will be kept clear to ensure there are no trip hazards.

Shorts, t-shirt and trainers should be worn for the testing session to allow access to the upper thighs.



WHEN SHOULD I REFUSE TO TAKE PART?

You are unable to participate as a subject in this study if any of the following criteria applies to you:

- Not a student at The University of Hertfordshire.
- Below the age of 18 years old or above 26 years old.
- Not active in exercise for a minimum of an hour each week.
- History of surgery to the lower limb.
- History of a knee injury in the past 12 months.
- Suffering from a musculoskeletal injury to the lower limb within the past 6 months.
- Suffering from any central or peripheral neurological conditions.
- Allergy to adhesive tape.
- Currently pregnant.

You can choose to withdraw at anytime without giving a reason and without prejudice from the tester.

WHAT ARE THE ADVERSE EFFECTS?

The tester will ensure the testing area is kept clear and tidy so there is a minimal risk of tripping. There is a small risk of getting muscle soreness after completing the exercises. The likelihood of this is small as there will be rest periods given between the exercises to ensure the muscles do not become fatigued. The tester will ensure that the technique is correct which again minimises any potential risks. The electrodes could potentially cause a minor skin allergic reaction but throughout the testing you will be monitored and if the skin becomes irritated, the tests will be ceased immediately. The likelihood of this occurring is very small as having an allergy to adhesive tape is within the exclusion criteria. If you are concerned about any of these risks, please don’t hesitate to contact the tester or supervisor for this study, Dr Andrew Mitchell.

CONSENT

Before the testing process begins, you will be asked to fill in a written consent form (LEC2). Before signing the form, you are encouraged to ask any questions you may have about the testing. Throughout the testing, any further questions can also be answered. Your participation in this study is voluntary and so you can discontinue the process at any time, without prejudice. If you choose to withdraw, any data already collected will not be used within the study.

PERSONAL DATA

Age, height and mass data will be collected to provide baseline comparative values between subjects. Your personal data and results will be kept anonymous; a subject number will be given so names will not be used at any point throughout the data analysis or within the write up of the study. All of your data will remain confidential in a password protected file on a password protected computer. Only the tester will have access to the data. Data will be deleted and destroyed at the completion of the assessment of the project, after the examination period is completed.

Appendix Four: Consent Form

LEC2 – This form is form is for use in Academic Year 2012 -13

UNIVERSITY OF HERTFORDSHIRE **Form LEC2**

SCHOOL OF LIFE SCIENCES ETHICS

CONSENT FORM

I, the undersigned, agree to take part in:

Approved Protocol Number or Registered Project Number ....LS1/10/12P........

Title of Study ...Electromyographic Analysis of Hip and Knee Exercises: A Continuum from Early Rehabilitation to Enhancing Performance...

to be carried out by

Name of Investigator(s) ....Megan Williams........

....

I confirm that the purpose of the study has been explained to me by the investigator and that I have been informed of the details of my involvement in the study.

I confirm that my questions regarding involvement with this study have been answered to my satisfaction.

I confirm that I understand that I am not obliged to participate in this study and that I may withdraw from the study at any stage without the need to justify my decision and without personal disadvantage.

Name of subject
(please print) ....

Signature of subject .... **Date** ....

Name(s) of investigator(s)
(please print) ....

....

Signature(s) of investigator(s) .... **Date** ....

.... **Date** ....

This form together with the Protocol Monitoring form (LEC5), must be returned to the School of Life Sciences Ethics Administrator on completion of the project work.

Appendix Five: Raw Data for Pilot Study One

The results for each task used to determine leg dominance. Three trials were performed, but this table shows the modal results for each subject.

Subject number	Kicking a ball	Hopping task	Regain balance	Writing hand
1	R	R	R	R
2	R	L	L	R
3	R	R	R	R
4	L	L	R	R
5	R	R	L	R
6	R	L	R	R
7	R	R	R	R
8	R	L	R	R
9	R	R	L	R
10	R	R	L	R
11	R	R	L	R
12	R	R	L	L
13	R	R	R	L
14	R	R	R	R
15	R	R	R	R
16	R	R	R	R
17	R	L	R	L
18	R	R	R	R
19	R	L	R	R
20	R	R	R	R
21	R	R	R	R
22	R	R	R	L
23	R	R	L	R
24	R	R	L	R
25	R	R	R	R
26	R	L	R	R
27	R	R	R	R
28	R	L	R	R
29	R	R	R	R
30	R	R	L	R
31	R	R	R	R
32	R	R	L	R
33	L	L	R	R
34	R	R	R	R
35	R	R	R	R
36	R	R	R	R
37	R	L	R	R
38	R	R	R	R
39	R	L	L	R
40	L	R	R	R
41	R	R	R	R
42	R	L	L	R
43	R	R	R	R

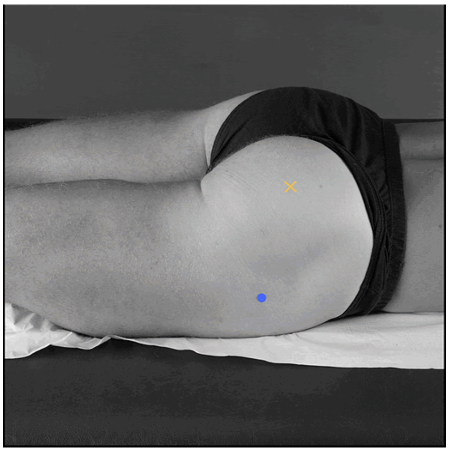
R: Right leg, L: Left leg

Appendix Six: Subject Characteristics for the Main Study

Subject number	Gender	Age (years)	Height (cm)	Mass (kg)
1	F	21	165.2	75.2
2	F	18	178.8	72.5
3	F	19	157.0	53.4
4	M	18	185.6	77.4
5	M	19	163.8	60.2
6	M	20	162.4	65.8
7	M	19	187.1	70.9
8	M	21	166.3	52.8
9	F	21	175.2	70.8
10	M	19	171.1	71.2
11	F	20	158.0	53.1
12	M	18	177.1	75.3
13	F	19	158.3	56.3
14	F	22	161.3	54.6
15	F	21	160.5	51.4
16	M	21	175.4	78.1
17	M	18	166.3	58.9
18	F	21	155.9	55.6
	Mean	**19.7**	**168.1**	**64.1**
	Standard Deviation	**1.3**	**9.7**	**9.8**

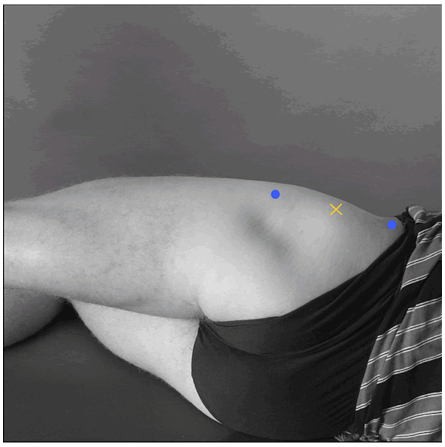
Appendix Seven: SENIAM Guidelines for EMG Electrode Placement

Gluteus Maximus



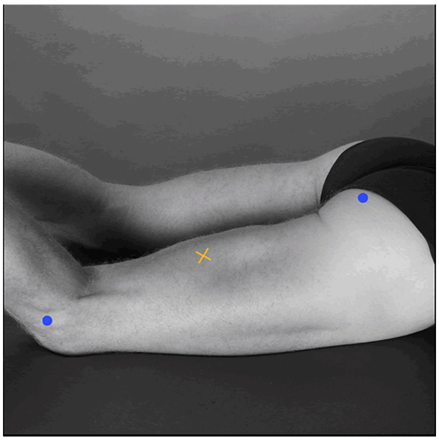
The GMax electrode was located 50% of the way between the sacrum and greater trochanter in the direction of the muscle fibres.

Gluteus Medius



The electrode for the GMed was placed half way along the line between the iliac crest and greater trochanter.

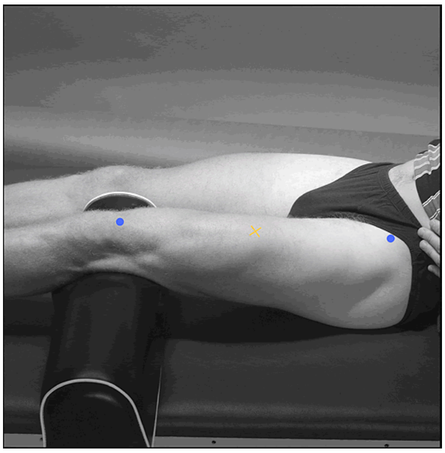
Biceps Femoris



For the BF, the electrode was positioned at 50% on the line between the lateral tibial epicondyle and the ischial tuberosity.

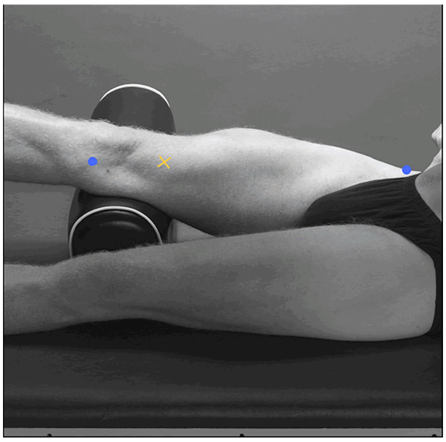
Rectus Femoris

The RF electrode was placed half way between the superior patella and the ASIS.



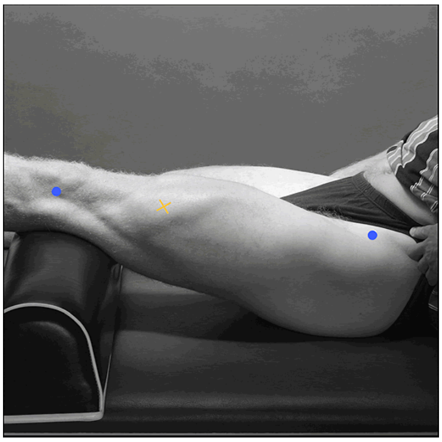
Vastus Medialis Oblique

Electrode placement for the VMO was 20% of the distance between the medial patella and ASIS, orientated perpendicular to the line between these landmarks.



Vastus Lateralis

The VL electrode was a third of the distance between the lateral patella and ASIS, orientated along the perceived line.



Appendix Eight: Knee Flexion Angle Results

Raw data for the knee flexion angles during the counter movement phase of the counter movement jump.

Trial 1	Trial 2	Trial 3	Average
102.4	105.0	112.6	106.7
89.5	95.5	91.5	92.2
77.2	84.9	86.9	83.0
77.1	86.4	95.5	86.3
64.7	63.7	61.2	63.2
94.5	94.5	88.2	92.4
113.4	112.8	104.1	110.1
96.7	100.1	100.3	99.0
91.6	100.2	106.5	99.4
96.0	91.8	97.3	95.0
67.5	70.2	68.9	68.9
79.2	75.9	75.6	76.9
77.1	74.8	77.1	76.3
88.3	88.3	86.3	87.6
64.9	73.5	71.2	69.9
68.2	66.2	69.3	67.9
82.6	83.5	82.5	82.9
85.9	87.0	86.6	86.5
		Mean	**85.7**
		Standard Error	**0.8**

Raw data for the knee flexion angles during the counter movement section of the single-leg vertical jump.

Trial 1	Trial 2	Trial 3	Average
80.9	79.3	76.7	79.0
89.0	90.0	86.7	88.6
60.1	60.2	60.6	60.3
66.8	69.3	66.8	67.6
55.8	55.6	55.6	55.7
75.8	71.1	72.3	73.1
59.4	65.2	63.1	62.6
60.9	62.5	59.8	61.1
65.3	67.0	64.4	65.6
67.1	63.7	67.2	66.0
67.5	64.2	60.1	63.9
60.6	59.8	67.6	62.7
60.3	63.0	67.0	63.4
68.0	69.7	68.4	68.7
51.6	53.8	52.7	52.7
48.2	56.5	52.6	52.4
59.8	57.0	64.3	60.4
61.0	62.1	61.2	61.4
		Mean	**64.7**
		Standard Error	**0.5**

Raw data for the knee flexion angles during the step up.

Trial 1	Trial 2	Trial 3	Average
74.9	81.1	81.0	79.0
86.0	86.0	86.9	86.3
87.4	85.9	88.0	87.1
78.1	77.1	76.8	77.3
70.5	61.6	71.8	68.0
85.3	83.4	89.3	86.0
63.5	64.0	67.9	65.1
85.0	85.7	87.2	86.0
79.5	84.9	79.5	81.3
63.9	63.8	67.6	65.1
86.4	85.9	88.7	87.0
70.9	67.3	73.1	70.4
85.8	83.6	89.1	86.2
64.0	63.4	67.2	64.9
85.1	74.9	84.2	81.4
86.2	85.9	87.3	86.5
69.9	69.4	73.9	71.1
		Mean	**78.2**
		Standard Error	**2.1**

Raw data for the knee flexion angles during the lateral step up.

Trial 1	Trial 2	Trial 3	Average
77.1	88.7	88.1	84.6
86.0	89.0	86.9	87.3
86.9	88.4	89.4	88.2
73.0	76.5	77.1	75.5
71.4	67.0	67.3	68.6
94.1	92.4	90.6	92.4
67.7	68.1	71.9	69.2
84.9	83.4	82.4	83.6
79.5	79.5	76.2	78.4
71.1	64.5	63.4	66.3
91.5	80.2	85.0	85.6
69.3	69.3	66.7	68.4
79.3	78.7	79.1	79.0
67.4	68.5	69.9	68.6
85.3	88.4	88.6	87.4
79.2	81.2	78.0	79.5
79.6	83.2	78.4	80.4
		Mean	**79.0**
		Standard Error	**2.4**

Raw data for the knee flexion angles during the step down.

Trial 1	Trial 2	Trial 3	Average
63.3	70.4	64.7	66.1
84.5	81.2	78.8	81.5
64.9	68.7	73.0	68.9
49.2	47.7	49.8	48.9
51.9	51.8	52.5	52.1
69.8	69.0	68.8	69.2
63.1	65.3	61.9	63.4
61.6	66.6	68.2	65.5
71.4	74.4	64.5	70.1
52.5	49.9	49.5	50.6
80.2	66.1	73.2	73.2
55.9	61.0	57.2	58.0
62.6	69.4	63.7	65.2
82.1	83.6	79.2	81.6
69.4	66.3	73.0	69.6
55.3	48.1	49.9	51.1
50.9	52.3	53.4	52.2
		Mean	**64.0**
		Standard Error	**3.0**