EMG MEASUREMENT OF THE ADDUCTOR MUSCLES DURING WALKING AND RUNNING

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Table of Contents

List of Tables

Chapter 5

Chapter 6

List of Figure

Chapter 3

Chapter 4

Chapter 5

Chapter 6

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Walaa Elsais

Glossary of Terms

AL	Adductor Longus
AM	Adductor Magnus
CAST	Calibrated Anatomical System Technique
CMC	Coefficient of Multiple Correlation
CPD	Contralateral Pelvic Drop
EMG	Electromyography
GMax	Gluteus Maximus
GMed	Gluteus Medius
Gr	Gracilis
LHam	Lateral Hamstrings
MGastro	Medial Gastrocnemius
MGastro	Lateral Gastrocnemius
MHam	Medial Hamstrings
MVIC	Maximal Isometric Voluntary Contraction
PCSA	Physiological Cross-Sectional Area
RF	Rectus Femoris
SD	Standard Deviation
SEM	Standard Error of The Measurement
SENIAM	Surface Electromyography for the Non-invasive Assessment of Muscles
TA	Tibialis Anterior
VLO	Vastus Lateralis Obliques
VMO	Vastus Medialis Obliques

Thesis Structure

Chapter 1

- **Overview and scope of the thesis**

Chapter 2

- **Literature review**

Chapter 3 (Study 1)

- **Using ultrasound to monitor the relative position of the adductor muscles in order to validate EMG measurement**

Chapter 4 (Study 2)

- **Consistency of EMG and kinematic variables in walking and running**

Chapter 5 (Study 3)

- **EMG profile for lower limb muscles during running**

Chapter 6 (Study 4)

- **Association of the frontal plane pelvic motion and the adductor activation pattern during running**

Chapter 7

- **Summary, Conclusions, Global Limitations and Recommendations for Future Work**

Abstract

Introduction

Increased pelvic drop has been linked to a range of musculoskeletal running injuries and may be linked to atypical activation pattern of the muscles surrounding the pelvis. However, to date, previous research investigating pelvic drop has focused on the abductor group, with minimal focus on the adductor group. Importantly, the over-activation of the adductor muscles could increase the adduction movement at the hip and therefore increase pelvic drop. However, this has not yet been investigated. Therefore, the studies within this thesis aimed to develop a valid and reliable protocol for measuring the activity of the adductor muscles and to investigate the association between adductor activation patterns and pelvic drop.

Methods

Ethical approvals were obtained from the University of Salford. **Study 1** quantified the relative movement of the adductor muscles under the skin at different hip joint angles and during incremental isometric contraction. In addition, it explored the relationship between adductor torque and the corresponding EMG amplitudes during ramped isometric contraction in 10 participants. **Study 2** investigated the between-day reliability for EMG measurements for the adductor muscles collected during both walking and running in 10 healthy runners. **Study 3** described the EMG profile and the inter-subject variability for the adductor muscles during running in 25 runners. **Study 4** investigated the association between the frontal pelvic plane movement and the adductor activation pattern during the early stance phase of running in 25 runners.

Results

The results of **Study 1** suggested that placing the surface electrodes centrally over the adductor muscles ensure that the adductor muscles remain within the EMG electrode detection volume. **Study 2** showed good to high between days reliability in both walking and running for the EMG that was developed for measuring adductor muscles activity. **Study 3** suggested that the adductor magnus and gracilis muscles activate at the foot strike while the adductor longus activates at toe off. **Study 4** showed that there was a significant strong positive correlation between the degree of adductor magnus activity and the pelvic drop angle.

Conclusions

The thesis establishes a robust and reliable method for measuring the activity of the adductor muscles using surface EMG electrodes in walking and running. Importantly, runners who exhibit increased pelvic drop also appear to demonstrate increased activity of adductor magnus during early stance phase. This finding motivates future clinical trials which could focus on muscle coordination retraining in order to improve kinematic patterns which have been linked to running-related injuries.

Chapter 1: Overview and scope of the thesis

1.1 Architecture of adductor muscles

The adductor muscle group is consisted of five separate muscles: the adductor magnus (AM), adductor longus (AL), adductor brevis, pectineus, and gracilis (Gr). As a group, the hip adductors make up 22.5% of the total muscle mass of the lower limb (Ito, 1996). This compares to a figure of 18.4% for the flexors, 14.9% for the abductors and 12.8% for the gluteus maximus (GMax) (Figure 1- 1). Similarly, the AM appears to have one of the largest physiological cross-sectional area (PCSA) of all lower extremity muscles, second only to the GMax (Ward, Eng, Smallwood, & Lieber, 2009). Additionally, the adductor group appears to have the second heaviest percentage (13.36%) of the total lower limb weight in the thigh region second only to the quadriceps group (21.19%) and larger than the hamstring group (9.59%) (Ito, Moriyama, Inokuchi, & Goto, 2003). Given the relative size of these muscles and corresponding capacity for generating muscle forces, it would seem intuitive that this muscle group plays an important role in human ambulation.
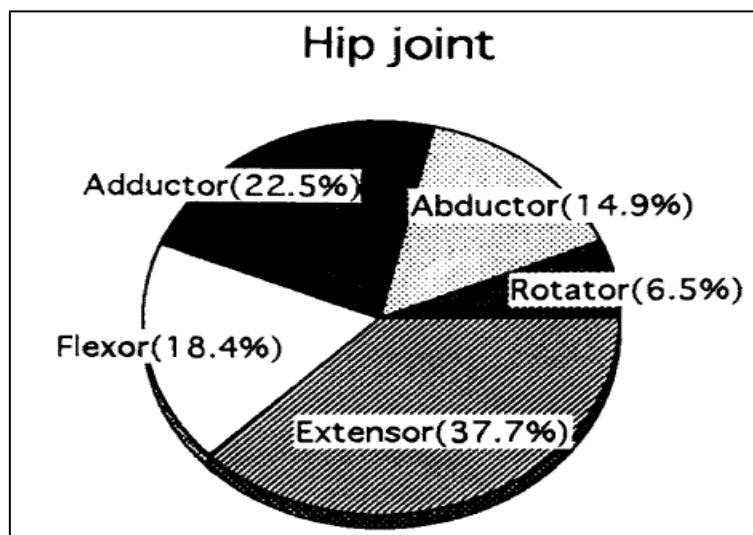
Figure 1- 1: The percent of mass of different hip muscle groups. Flexor group: sartorius, rectus femoris, iliopsoas. Extensor group: gluteus maximus, biceps femoris, semitendinosus, semimembranosus. Adductor group: gracilis, pectineus, adductor longus, adductor brevis, adductor magnus. Abductor: piriformis, gluteus medius, gluteus minimus. Rotator: quadratus femoris, obturatorius externus, gemellus, obturatorius internus.

The 'moment arm' of a muscle determines the direction and size of the moment that it can exert at a given joint (Levangie, 2011). Although all of the hip adductors exhibit relatively large moment arms for frontal plane moments, different hip adductors also exhibit moment arms of varying magnitudes in the sagittal plane. For example, the AM muscle is actually a very important hip extensor when the hip is in the flexed position. This is because the AM, at 90° hip flexion, has a longer moment arm for hip extension compared the main hip extensor muscle (Nemeth & Ohlsen, 1985). Similarly, the AL acts to produce hip flexion during certain sagittal plane angles (Dostal, Soderberg, & Andrews, 1986). The variations in muscle moment arm length can reflect the muscle's capacity to generate force. For instance, the force generated by the AM is approximately 70% of that produced by the GMax. Also, it represents about 83% of the force produced by the hip abductors (Arnold, Ward, Lieber, & Delp, 2010).

Taken together, these anatomical studies demonstrate that the adductor muscles have the capacity to produce large joint moments in both the frontal and sagittal planes. Given this capacity, it is important to understand the precise function of the adductor muscles during human gait and whether dysfunction of these muscles could be associated with gait dysfunction and/or musculoskeletal pathology.

1.2 The lack of research measuring adductors in walking and running

In order to understand the breadth of previous research investigating the function of the adductors, a systematic search of the literature was performed. The search strategy focused on electromyography (EMG), the primary technique used for measuring muscle function. The precise search strategy is documented in Appendix I and the results shown in Figure 1- 2). Approximately 70 papers were identified and retrieved for further analysis. This number of studies is quite small when compared with those addressing other muscles. For example, for

the hip extensor, a quick search was made and identified approximately a thousand articles. Among these 70 articles, only 11 papers investigated EMG activity for the adductor muscles during gait. In addition, only eight articles exploring the EMG activity for adductor muscles in musculoskeletal disorders were identified. These articles will be discussed in more detail in the following paragraphs.
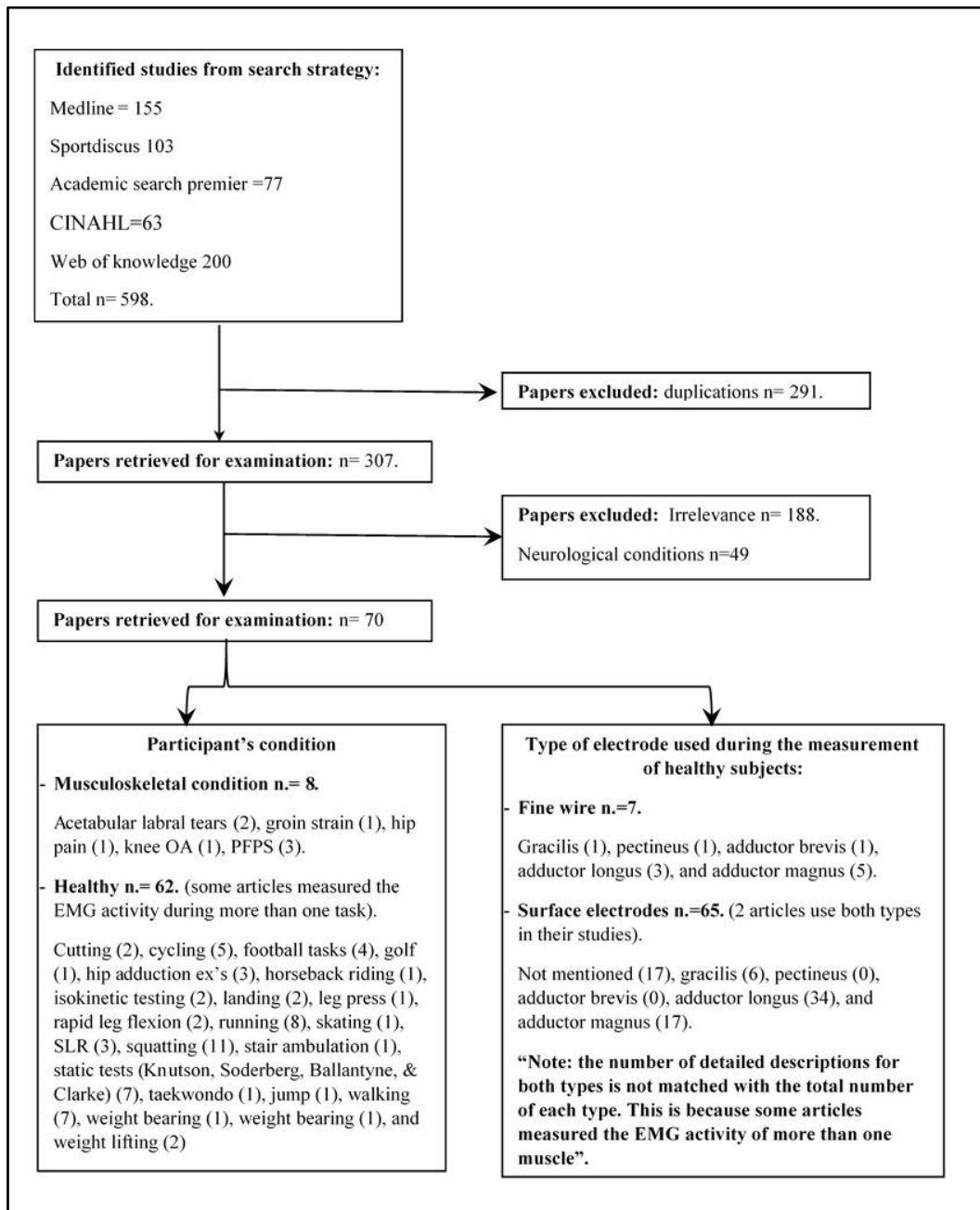
Figure 1- 2: Diagram of paper identification strategy.

A small number of studies have explored EMG activity for the adductor muscles during gait, including three studies of walking and four of running. In the walking articles, Bogey and Barnes (2016) proposed that the hip adductors did not act as a single synergistic group. The AM synergistically assisted other hip extensors and produced forces that were out-of-phase with the other hip adductor forces. Moreover, Lee and Hidler (2008) reported higher activity of the AL in the early and mid-swing phases of overground walking compared to their activity during treadmill walking. Similarly, an increased level of activity for the AL was observed as walking speed increased (Hu et al., 2010). These studies used an invasive approach, fine wire technique (Bogey & Barnes, 2016), did not specify the recording method (Lee & Hidler, 2008), or only mentioned a method of application which is not exist for this group of muscles such as the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) recommendations (Hu et al., 2010).

Similarly, four studies investigated the activity of the adductor muscles during running. Montgomery, Pink, and Perry (1994) proposed that the AM muscle afforded pelvic stabilisation and assisted the main hip extensors in the early stance phase. Three studies investigated the effect of specific exercise programmes on EMG activity of the adductor muscles. For example, a higher EMG activity for the AL was reported during deep water running compared to level and water walking (Kaneda, Sato, Wakabayashi, & Nomura, 2009). Similarly, an increased EMG amplitude of the adductor muscle during the preparatory phase of landing during running tasks was reported after a period of plyometric training (Chimera, Swanik, Swanik, & Straub, 2004). Finally, the level of EMG adductor activity was shown to be influenced by the percentage of body weight support while running on a positive-pressure treadmill (Hunter, Seeley, Hopkins, Carr, & Franson, 2014). Again, these studies used either fine wire (Montgomery et al., 1994), which may not be appropriate for the adductor muscles, especially during dynamic tasks, or a placed surface EMG electrodes using

a manual palpation technique (Chimera et al., 2004; Hunter et al., 2014; Kaneda et al., 2009) in order to locate this group of muscles. The palpation manual technique of application could result in the inappropriate placement of electrodes. Accordingly, it could lead to crosstalk especially in this relatively small area.

Finally, four articles investigated the EMG activity of the adductor muscles during both walking and running activities. Mann, Moran, and Dougherty (1986) proposed that the AL was silent in the stance phase of walking, jogging, and running, and was active in the late toe-off only. In addition, the activity of the hip adductor muscles was similar between genders during walk and run conditions (Chumanov, Wall-Scheffler, & Heiderscheit, 2008). Moreover, Wall-Scheffler, Chumanov, Steudel-Numbers, and Heiderscheit (2010) concluded that the adductor muscle is speed dependent (i.e. as the speed increases the muscle activity increases). In contrast, Tsuji, Ishida, Oba, Ueki, and Fujihashi (2015) found that the activities of the hip adductors did not significantly differ among all speeds. There is minimal information on the biomechanical function of the adductor muscle groups during gait. As explained earlier, the adductor muscles have capacity to generate joint moments in both the sagittal and frontal plane. However, it is not clear when, and to what degree, these muscles are active during walking and running and how such action may contribute to sagittal and frontal joint moments during gait. Muscle function may also be characterised by maximal force testing (e.g. isometric, isokinetic) and this may provide further insight into the force generating capacity of these muscles during gait. Given this lack of understanding, considerably more research is needed to understand the role of this group of muscles during human gait.

1.3 Adductor activity in musculoskeletal disorders

It has been suggested that altered frontal plane motion of the pelvis could be associated with gait dysfunction, poor gait stability during walking, and hip and knee pathology during running (Dean, Alexander, & Kuo, 2007; McAndrew, Wilken, & Dingwell, 2011). However, there is minimal understanding of the links between abnormal pelvic motions and abnormal adductor coordination. Moreover, there have been only a small number of studies investigating adductor EMG patterns in musculoskeletal conditions and the majority of these studies have focused on the effects of an intervention (i.e. they aim to explore the effect of certain variables on this group) (Aminaka, Pietrosimone, Armstrong, Meszaros, & Gribble, 2011; Brandt et al., 2013; Cerny, 1995; Dwyer, Lewis, Hanmer, & McCarthy, 2016; Glaviano & Saliba, 2016; Krebs, Robbins, Lavine, & Mann, 1998; Morrissey et al., 2012; Solomonow-Avnon et al., 2016). For instance, neither the exercise programme nor patellar tapping influenced the level of activity for the AM muscle in patellofemoral pain (PFP) patients (Cerny, 1995). Similarly, there was no significant difference in AL activity during ascending and descending phases of lounge exercise in patients with acetabular labral tears. Compared to asymptomatic participants, subjects with PFP displayed an earlier onset and longer activation of the AL during stair-climbing activity (Aminaka et al., 2011). To date, it is unclear exactly how this group of muscles behaves during gait in participants with musculoskeletal disorders.

1.4 Difficulties in measuring adductor muscles activity

It is possible that the paucity of research investigating the function of the adductors during walking and running is due to the difficulties in measuring this muscle group. For example, there are no clear guidelines for surface EMG electrode placement for the adductor muscle group. Moreover, the adductor muscles are very close to each other at the upper medial side

of the thigh. Thus, small errors in electrode placement using manual palpation, for example, could increase the possibility of cross talk (Watanabe, Katayama, Ishida, & Akima, 2009). Some researchers have attempted to use fine wire electrodes, which are inserted directly into the muscle. (Mann et al., 1986; Montgomery et al., 1994). However, this invasive technique may not be appropriate for the adductor muscles, especially during dynamic tasks.

Although the adductor muscles are some of the largest muscles in the lower limbs and have a large capacity for generating moments in both the sagittal and frontal planes, there has been very little study into the biomechanical function of this group. Consequently, little is known about the precise role of the adductors during gait. Therefore, the aim of this thesis is to develop a standard and reliable method for measuring this group and to understand typical activation patterns during running gait. With this methodology and understanding of function, it was then sought to explore how patterns of frontal plane motions in running, previously linked to pathology, could be related to adductor function.

Specifically, the aims of this project were to:

Chapter 3:

i.   Establish a standardised method to measure the activity of the adductor muscles activity using surface EMG electrodes. This was achieved by answering the following research questions:

- *Experiment 1*: Is the position of the adductor muscles, relative to the skin, similar at various hip joint angles in standing position?

- *Experiment 2*: Is the position of the adductor muscles, relative to the skin, similar while performing ramped isometric contraction?

- *Experiment 3*: Is there a relationship between torque produced by adductor muscles and the magnitude of the corresponding EMG signal during isometric contraction?

Chapter 4:

ii.   Investigate the degree of consistency between EMG measurements from the adductor muscles during both overground walking and running activities. This was achieved answering the following research questions:

- Do healthy subjects demonstrate consistent EMG patterns during walking in repeated sessions, separated by approximately 1 week?
- Do healthy subjects demonstrate consistent EMG patterns during running in repeated sessions, separated by approximately 1 week?

Chapter 5:

iii.   Describe the typical EMG profile for the adductor muscles during running as well as the inter-subject variability.

iv.   Describe the typical EMG profile for the major lower limb muscles collected during running as well as the associated inter-subject variability.

Chapter 6:

v.   Investigate the association between the frontal plane movement and the EMG activity of the hip adductor muscles at specific point of the stance phase of running. This was achieved by answering the following research question:

- Do EMG measures of adductor function associate with lower limb frontal plane kinematics during the early stance phase of running.

Chapter 2: Literature review

This chapter presents a review of the available literature on the adductor muscles. Firstly, the functional anatomy of this group will be explained. Secondly, the basics of EMG measurements and the factors that can influence their validity and how the previous studies processed and normalised EMG signals will be described. Additionally, an overview of reliability studies on walking and running were presented. Furthermore, the typical profile for the adductor muscles presented in the previous literature will be discussed. Finally, the issue of their abnormal activation patterns and how this could lead to running related injuries will be addressed.

2.1 Anatomy and architecture of adductor muscles
2.1.1 Functional anatomy of the adductor muscles

The most important function of the hip muscles appears during weight-bearing activities. In weight-bearing activities, the muscles function to move or support the head, arm, and trunk (approximately two-thirds of the body weight) (Levangie, 2011). Consequently, the hip muscles acclimatise their structure to the required role. In addition, the alignment of the hip joint muscles and the large range of motion offered by the hip joint means that the muscles have a role in more than one movement plane. For example, the adductor muscles may act as hip flexors in the neutral hip joint but as hip extensors when the hip joint is already flexed (Basmajian & De Luca, 1985). This example explains how the adductor muscles function not only in the frontal plane but also in other movement planes.

2.1.1.1 Anatomy of the adductor group

The hip adductor muscle group generally consists of five muscles: the pectineus, adductor brevis, AL, AM, and Gr muscles. The hip adductor muscles are situated in the anteromedial aspect of the thigh. The adductor brevis, AL, and AM muscles originate from the inferior

ramus of the pubis and insert along the linea aspera of the femur. The Gr muscle is the only two-joint muscle in the adductor group. It originates on the symphysis pubis and pubic arch and inserts on the upper medial surface of the shaft of the tibia. The contribution of the adductor muscles to hip joint function is a major point of debate. With excessive femoral anteversion, the moment arms of the adductor brevis, pectineus, and the middle fibres of AM muscles switched from medial rotational to lateral rotational lines of pull. After examining the changes in the moment arms with femoral anteversion or combined hip medial rotation and knee flexion, Arnold and Delp (2001) concluded that the adductors were unlikely to have a strong influence on the medially rotated hip position during the gait cycle.

The AM muscle is the biggest member of the hip adductor group in terms of muscle mass. It is a large triangular muscle, situated on the medial side of the thigh. The AM is split into four parts based on courses of the corresponding perforating arteries from the deep femoral artery (Figure 2- 1). The first portion of the adductor magnus (AM1) is the part of the muscle proximal to the first perforating artery and next to it, the second portion (AM2) is located. Distal to AM2, the third portion, AM3, is situated but proximal and lateral to the adductor hiatus. The final portion, AM4, is located distal and medial to the adductor hiatus. The AM is divided into a hamstrings part and adductor part. The hamstrings part corresponds to AM4 that attaches to the adductor tubercle at the distal end of the femur, while the adductor part corresponds to the remaining three portions attached to the linea aspera of the femur. The AM1 portion is designed for stabilising the hip joint. Because of the presence of longer muscle fibres than the AM1, the other three portions function as displacers for moving the thigh through a large range of motion. Each portion of the AM may have its own role depending on its dynamic circumstances (Takizawa, Suzuki, Ito, Fujimiya, & Uchiyama, 2014). Although the role of the adductor muscles may be less clear than that of other hip muscle groups, the relative importance of the adductors should not be underestimated.
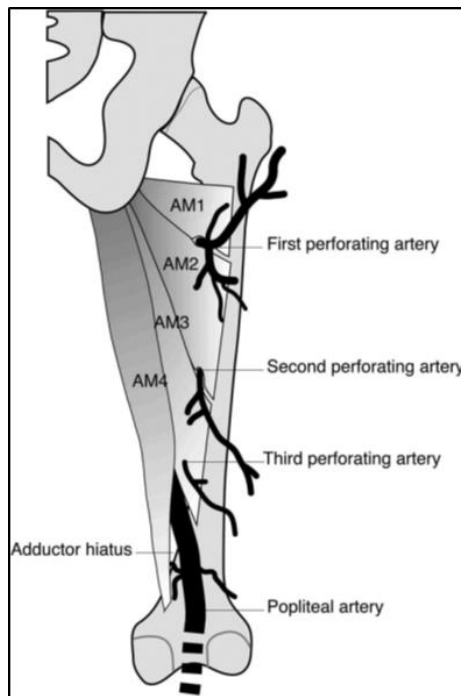
Figure 2- 1: Posterior view the four portions of the adductor magnus (AM1-AM4) based on courses of the corresponding perforating arteries from the deep femoral artery (Arnold et al., 2010).

2.1.1.2 Anatomy of the hip flexor group

Nine muscles have action lines crossing the hip joint anteriorly. Of these, the primary hip flexors are the iliopsoas, rectus femoris (RF), tensor fascia lata, and sartorius. The iliopsoas muscle is considered the main hip flexor. It comprises of two separate muscles, the iliacus muscle and the psoas major muscle, both are attached to the femur by a common tendon. The primary function of the hip flexors during gait is to move the swinging limb forward. In addition, they withstand the strong hip extension forces that occur during the early stance. The secondary hip flexors, the pectineus, AL, AM, and the Gr muscles, are mainly adductors of the hip joint (Figure 2- 2). Each, however, is capable of contributing to hip joint flexion during gait, but that contribution is dependent on hip joint position (Levangie, 2011).
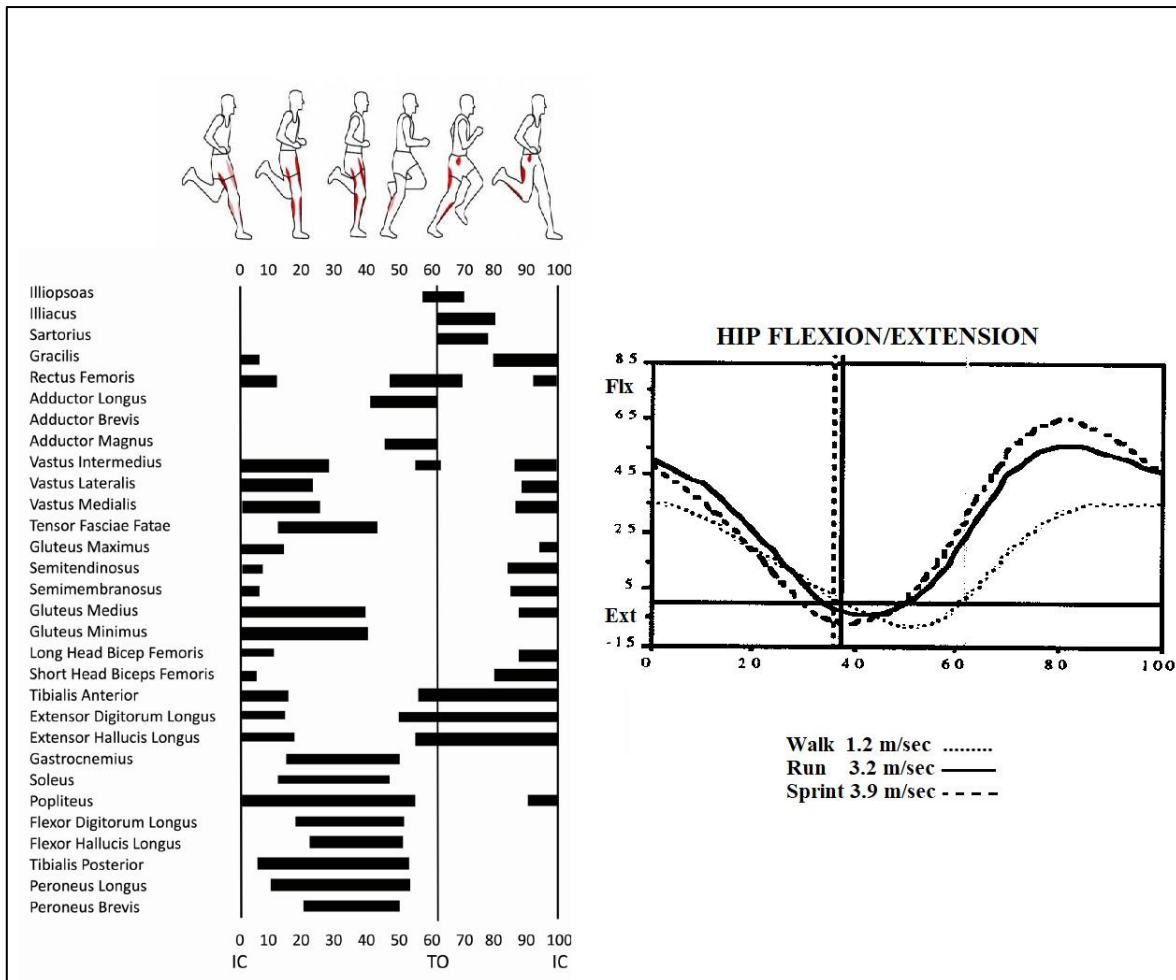
Figure 2- 2: Timing of various lower limb muscles during running activity and sagittal plan hip kinematics. Image adopted and amended from (Kamen & Gabriel, 2010; Novacheck, 1998).

2.1.1.3 Anatomy of the hip extensor group

The GMax and the hamstring muscle group are function primarily as hip extensors. The GMax is a one-joint muscle that originates from the posterior sacrum, dorsal sacroiliac ligaments, sacrotuberous ligament, and a small portion of the ilium. Its superior fibres insert into the iliotibial band while its inferior fibres insert into the gluteal tuberosity. The maximus activates primarily against a resistance greater than the limb weight. In addition, the posterior fibres of the gluteus medius (GMed), the posterior portion of the AM muscle, and the piriformis muscle may assist the maximus in its function. The other primary hip extensor, the hamstring, comprises of three two-joint extensor muscles: the long head of the biceps femoris, the semitendinosus, and the semimembranosus muscles. Each of these muscles

12

originate from the ischial tuberosity. The biceps femoris moves across the posterior aspect of the femur to insert into the fibular head and lateral aspect of the lateral tibial condyle. The other two hamstrings insert into the medial aspect of the tibia. All three parts of the hamstring muscles support the extension of the hip in the early stance as well as serve as the main knee flexors (Levangie, 2011).

2.1.1.4 Anatomy of the hip abductor group

The GMed and the gluteus minimus muscles are function primarily to abduct the hip joint. The GMed originates from the lateral surface of the ilium and inserts into the greater trochanter, beneath the GMax (Robertson et al., 2008). It has three parts (anterior, middle, and posterior) that function asynchronously during movement at the hip (Soderberg & Dostal, 1978). The superior fibres of the GMax and the sartorius muscles may help the abduction of the hip against the strong resistance. The tensor fascia lata muscle is given variable credit for its contribution and may be effective as an abductor only during simultaneous hip flexion. The GMed and gluteus minimus muscles function together either to abduct the femur (as in open kinetic chain movement i.e. the distal lever moves in space) or, more importantly, to stabilise the pelvis and superimposed head, arm and trunk (HAT) in unilateral stance against the effects of gravity. Consequently, the gluteus minimus and GMed muscles will offset the gravitation adduction torque on the pelvis (pelvis drop) around the weight-bearing hip in unilateral stance (Levangie, 2011).

This anatomical background demonstrates that the hip muscles are integrated muscle groups and they may function together in certain situations. In addition, the adductor muscles are important members in hip musculature and they have the capacity to play a role in the frontal and other movement planes. Compared to other hip muscles, the role of the adductor muscles may be unclear during gait and has not previously been investigated in detail. Therefore, it is

important to understand the precise function of the adductor muscles during human ambulation

2.1.2 Morphological structure of the adductor muscles

Muscle architecture is a primary determinant of muscle function. In addition, architectural differences between muscles are the best predictors of force generation capacity (Lieber & Fridén, 2000). Therefore, understanding this structure–function relationship is of great importance. Among the hip muscles, the hip adductor group contributes to 22.5% of the total muscle mass of the lower extremity (Ito, 1996). Moreover, they take up 27% of the mass of the thigh musculature (Takizawa et al., 2014). Although the AM is considered to have the second largest PCSA among the lower limb muscles, it has the capacity to generate extension force equal to the force produced by the GMax (Ito et al., 2003). Similarly, the summation of PCSA for the adductor muscles is larger than the PCSA of the GMed muscle (Williams, Wilson, Daynes, Peckham, & Payne, 2008). However, although they have a large muscle mass of the lower limb, the contribution of the adductor muscles during gait still unclear.

2.1.3 Factors affecting the force generation of the adductor muscles

Many factors influence the capacity of a muscle to generate forces. Among these factors, muscle architecture appears to be a critical factor (Lieber & Fridén, 2000). The common parameters included in an architectural analysis are PCSA and pennation angle. PCSA is a measure of the muscle cross section perpendicular to the fibres, while pennation angle is the fibre angle in relation to the force-generating axis. In addition, the PCSA is influenced by the angle of pennation of the muscle fibre while, the pennation angle is affect by the range of motion. Thus, the capacity to generate force is influenced by the PCSA and angle of pennation (Williams et al., 2008). Despite the fact that the pennation angles of the adductor muscles are smaller than those of the GMax and GMed, they are capable of generating a

higher force than these two muscles (Arnold et al., 2010; Ward et al., 2009). In addition, the dynamic role of the limb can be deeply understood by obtaining measures of muscle moment arms. Moment arms can transform the linear forces developed by the muscles into rotational moments that result in joint movements (Williams et al., 2008).

The flexion-extension moment arms of the major pelvic muscles at the hip have been estimated in previous studies using a modelling approach (Arnold et al., 2010; Dostal et al., 1986; Nemeth & Ohlsen, 1985; Ward, Winters, & Blemker, 2010). Moment arms were scaled by segment length and presented as muscles acting to extend or flex the hip joint. For example, the moment arm length of the AM muscle is twice that of GMax at 90° of hip flexion but it decreases in the anatomical position (at $0^°$ hip angle) (Nemeth & Ohlsen, 1985). This makes the hamstrings and GMax more effective hip extensors than the AM when the hip is extended, as observed in the anatomical position (Dostal et al., 1986; Nemeth & Ohlsen, 1985). Despite the fact that GMax has a larger PCSA, similar maximum extension torques have been recorded for the GMax and AM in the sagittal plane (Ward et al., 2010).This is due to the fact that the AM has a larger hip extension moment arm when the hip is flexed. However, the AL produces hip flexion with a moment arm length nearly equal to that of the sartorius and RF and longer than that of the iliopsoas (Dostal et al., 1986). Changes in the moment arm length can influence the force produced by the muscles. The AM produces a force of about 70% of that of the GMax, and the hip adductor muscles produce a force of about 83% of that of the hip abductors (Arnold et al., 2010). Consequently, the adductor muscles can produce significant joint moments both in the frontal and sagittal planes.

The morphological structure of the adductor muscles reveals that a huge muscle mass occupies the whole medial aspect of thigh. In addition, they have a large PCSA and relatively small pennation angle as compared to the other lower limb muscles. Furthermore, they have

the capacity to produce a force that is higher than that produced by the other hip

musculatures. Therefore, further research is required to understand the role the adductors play

during ambulation and whether dysfunction of these muscles could play a role in gait

dysfunction and/or musculoskeletal pathology. Accordingly, kinesiological EMG can give us

a better understanding of how this group of muscles behaves during human gait. Therefore, in

the following section, the background of EMG, how the signals originate and propagate will

be discussed. In addition, the process of collecting high-quality signals and processing and

normalising the acquired data will be explained. Finally, the limitations of EMG

measurement will be highlighted.

2.2 EMG measurement of skeletal muscle
2.2.1 Background and history of EMG.

Electromyography is an electrodiagnostic method interested in developing, recording, and

analysis of myoelectric signals (Kamen, 2004). Such signals are created by physiological

differences in the state of muscle fibre membranes (Basmajian & De Luca, 1985). Unlike the

classical neurological EMG, where an artificial muscle response due to external electrical

stimulation is analysed in static conditions, the focus of kinesiological EMG on studying of

the voluntary neuromuscular activation of muscles within postural tasks, functional

movements, work conditions, and treatment/training regimes. Nowadays, it remains one of

the only direct windows into the neural codes that produce muscular contraction, force, and

movement.

2.2.2 Motor unit action potential
The origin of the EMG signals

Understanding EMG signal generation in human muscles is helpful for understanding the

principles of obtaining a valid measurement. The motor unit is considered as the smallest

single functional unit that can describe neural control of muscle contraction. The term unit

expresses the behaviour that all muscle fibres of a given motor unit act "as one" within the innervation process. The single motor unit includes one anterior horn cell, one axon, its neuromuscular junctions, and all the muscle fibres innervated by this axon (Figure 2-3) (Enoka, 2015; O'Sullivan, Schmitz, & Fulk, 2013). Once activated, the motor neuron spreads the action potential (MUAP) along its axons. When the MUAP accomplishes a synaptic end bulb, it activates a sequence of electrochemical events to release the neurotransmitter. This chemical transmitter moves from the synaptic cleft to bind with the receptors located in the motor end plate of the muscle, which consists of neurotransmitter receptors (Konrad, 2005). Activation of these receptors results in an influx of sodium ions into the cell membrane of the muscle and efflux of potassium ions out of the cell membrane leading to a depolarisation of the postsynaptic membrane, causing an action potential (Figure 2-4).
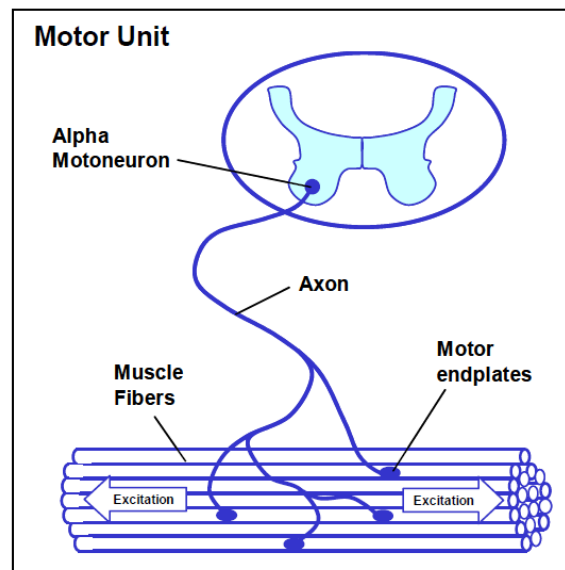
Figure 2-3: Motor unit structure (Konrad, 2005).

Two voltage-gated channels play a role in the action potential process. Initially, the voltage-gated sodium ion channels open to allow the influx of the sodium ion inside the membrane, causing membrane depolarisation. In this stage, the negative membrane potential changes into positive (up to +30 mV) for about 2 ms. Thereafter, the potassium gates open to permit the efflux of potassium ions. In a few milliseconds, the membrane permeability is revered to

restore the resting membrane condition in a process called repolarisation (Tortora & Derrickson, 2008). In the resting status, the ionic balance is kept by an active sodium-potassium pump, forming a resting potential of approximately -80 to -90 mV (Tortora & Derrickson, 2008). When the action potential is generated, there is a transient negative shift from the original resting membrane potential, and this is referred to as the after-hyperpolarisation stage (Villalobos, Foehring, Lee, & Andrade, 2011).
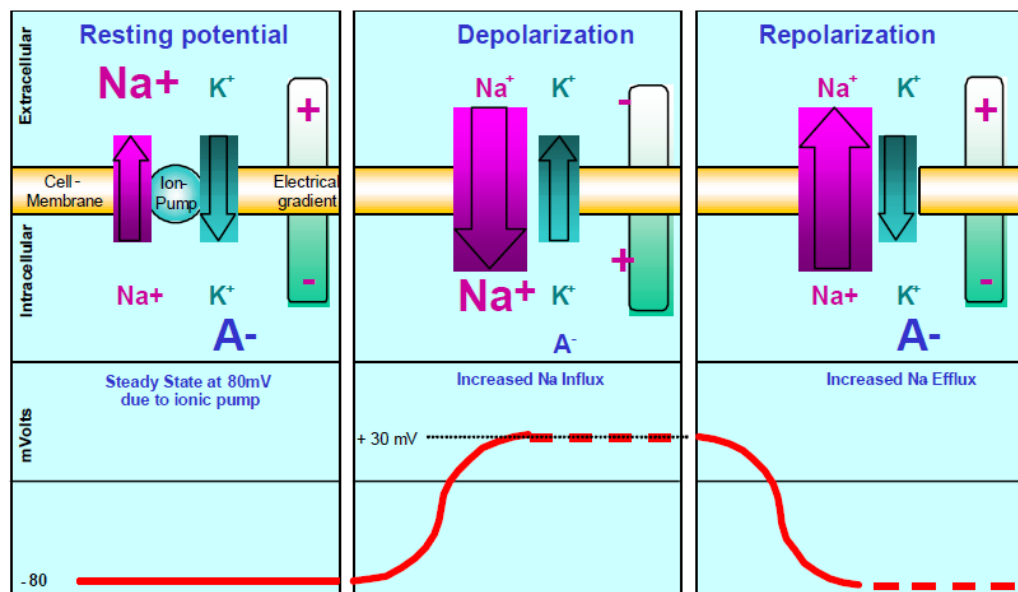
Figure 2-4: Depolarisation/repolarisation process within the excitable membrane (Konrad, 2005).

The action potential travels along the length of the muscle fibre at a speed of between 2 and 6 m.s^{-1} (Baker, 2013). Although this takes a few milliseconds, it is not likely that any other action potential occurs. The generated action potential propagates along the muscle fibre membrane and provokes the release of calcium ions into the sarcoplasm. Thereafter, the calcium ions attach to the actin myofilaments (contractile elements of skeletal muscle), causing a sliding of these myofilaments, and, finally, the whole muscle shortens. The sliding of the myofilaments is energised by ATP. The longer axon of the motor neurons may need a longer time for activation. The aforementioned mechanism occurs in any type of muscle

contraction in healthy individuals (Tortora & Derrickson, 2008). This type of potential is influenced by the type of training and the type of muscle fibre: slow or fast-twitch fibres (Hammelsbeck & Rathmayer, 1989; Kamen & Gabriel, 2010; Moss, Raven, Knochel, Peckham, & Blachley, 1983). All these variations in membrane potential, depolarisation, overshoot, repolarisation, and after-hyperpolarisation, are known to as the action potential (AP) process (Figure 2-5).
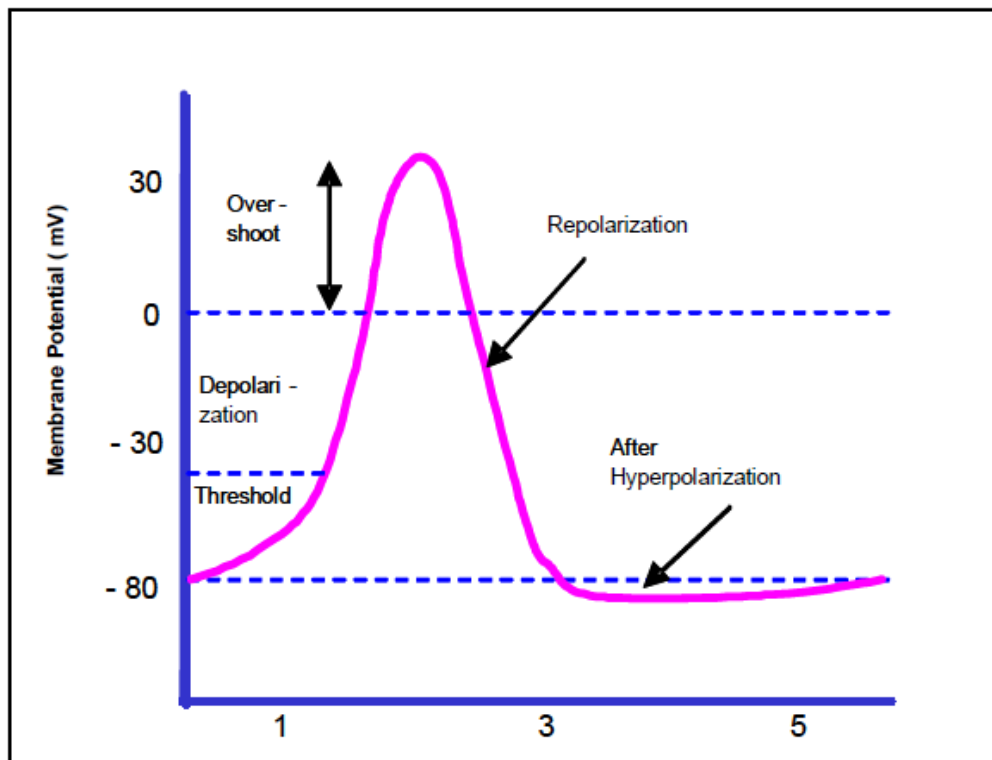
Figure 2-5: Generation of action potential.

Motor unit action potential

The term innervation ratio is used to express the number of muscle fibres innervated by a motor neuron (Kamen, 2004). The motor unit action potential (MUAP) is the summation of the electrical activity of all muscle fibres activated within the motor unit, which is proportional to the innervation ratio (Henneman, Somjen, & Carpenter, 1965). Therefore, as the number of recruited motor units increases, the strength of the contraction increases. Furthermore, a higher activation frequency produces a higher muscular force (Robertson,

2014). A force-time response to an action potential is a twitch. It is relatively common for a motor unit to receive a number of action potentials, resulting in overlapping twitches (Enoka, 2015). This may lead to oscillating forces when it is insufficient to initiate an action potential. Moreover, doublets (two short bursts of motor unit activation) and synchronisation (more than one motor unit fires at the same time) are important processes that can alter the muscular force (Clamann & Schelhorn, 1988). In summary, the EMG signal is the electrical summation of the activity of all active motor units in the detecting volume. These signals consist of the amplitude of both negative and positive components. This amplitude reflects the intensity of the muscular contraction; however, the relationship between amplitude and muscular force is frequently non-linear (Solomonow, Baratta, Shoji, & D'Ambrosia, 1990).

2.2.3 Factors that influence the validity of EMG measurement

A number of factors have an impact on the quality of the EMG signals. Farina, Merletti, and Enoka (2004) categorised and summarised these factors into two main groups: nonphysiological and physiological (Table 2- 1). Some of these effects are not intuitive and differ with experimental conditions. Nonetheless, useful information can be gained from the surface EMG, especially when the experimental protocol minimise the effect of these factors . Proper electrode placement could help in reducing of some of these factors. On the other hand, there are some other factors which could influence the EMG signal quality which need to be considered carefully during measurement of the adductor muscles. These factors include soft tissue characteristics and muscle movement in relation to the skin, during either dynamic movement or muscle contraction and crosstalk. These factors are integrated and could lead to inaccuracy of the EMG measurement. Moreover, these factors can greatly differ from individual to individual (and even within individual) and prohibit a direct quantitative comparison of EMG amplitude parameters calculated using unprocessed EMG signals.

Table 2- 1: Factors influencing the surface EMG signals

Non-physiological	Anatomic	• Shape of the volume conductor • Thickness of the subcutaneous tissue layers • Tissue inhomogeneities • Distribution of the motor unit territories in the muscle • Size of the motor unit territories • Distribution and number of fibres in the motor unit territories • Length of the fibres • Spread of the endplates and tendon junctions within the motor units • Spread of the innervation zones and tendon regions among motor units
	Detection system	• Skin-electrode contact (impedance, noise) • Spatial filter for signal detection • Interelectrode distance • Electrode size and shape • Inclination of the detection system relative to muscle fibre orientation • Location of the electrodes over the muscle
	Geometrical	• Muscle fibre shortening • Shift of the muscle relative to the detection system
	Physical	• Conductivities of the tissues • Amount of crosstalk from nearby muscles
Physiological	Fibre membrane properties	• Average muscle fibre conduction velocity • Distribution of motor unit conduction velocities • Distribution of conduction velocities of the fibres within the motor units • Shape of the intracellular action potentials
	Motor unit properties	Number of recruited motor units Distribution of motor unit discharge rates Statistics and coefficient of variation for discharge rate Motor unit synchronization

Soft tissue characteristics

The magnitude of the EMG signal recorded at the surface of the skin is strongly influenced by the electrical conductivity of the tissues between the muscle and the electrode. A fatty layer between the electrode and the muscle can be the cause of inaccuracy in the results of the measurement obtained with the application of surface EMG (Farina & Mesin, 2005; Nordander et al., 2003). Kuiken, Lowery, and Stoykov (2003) state that an increase in the subcutaneous fatty layer leads to a decrease in the EMG signal amplitude and an increase in

the likelihood of crosstalk from the adjacent muscles. Similarly, Farina and Rainoldi (1999) suggest that the subcutaneous tissue layers attenuate the potential distribution present at the muscle surface. The percentage of fat tissue layer could be a problem especially in the inner thigh region and could influence the amplitude of the EMG signals.

The reason behind the decrease in EMG amplitude is that the presence of subcutaneous fatty tissues reduces the spectrum of the EMG signal. The detection volume for the surface EMG electrodes is within 10–20 mm from the surface layer, i.e. the skin (Barkhaus & Nandedkar, 1994; Fuglevand, Winter, Patla, & Stashuk, 1992; Winter, Fuglevand, & Archer, 1994). The anatomical dissimilarities such as the percentage of subcutaneous fat mislead the data interpretation, particularly when comparing among individuals. In addition, skinfold thickness is negatively correlated with motor unit yields during voluntary contractions (Zaheer, Roy, & De Luca, 2012). Therefore, the subcutaneous tissue between the muscle and electrode must be taken into consideration during EMG measurements (De la Barrera & Milner, 1994; Herda et al., 2010; Petrofsky, 2008).

Muscle movement in relation to the skin during dynamic tasks

The relative motion of muscles under the skin can affect the process of EMG measurement. For example, Enoka (2015) stated that the muscles move slightly as they contract as a result of changes in the joint angle. The contraction of contractile elements of a muscle causes a transverse displacement of its fibres (i.e. the fibres move parallel to the skin) in relation to the overlying skin. This displacement occurs as a result of muscle shortening during contraction. Therefore, any significant difference in the muscle dimensions will affect the relative electrode location and, as a result, the detection volume of the EMG electrode changes. However, there is no study quantifying the adductor muscles movement relative to the skin at different hip joint angles.

Muscle movement in relation to the skin during contraction

Similar to muscle movement during a dynamic task, muscles move in relation to the overlying skin as a result of muscle contraction. Delaney, Worsley, Warner, Taylor, and Stokes (2010) examined the contractile capacity of the quadriceps muscle using ultrasound imaging technique during submaximal and maximal voluntary contractions. They found significant changes in the width of the RF (total distance between muscle borders) during incremental muscle contraction of the quadriceps muscle. Likewise, Rainoldi et al. (2000) assessed the geometrical artefacts on the EMG signals for the vastus medialis obliques and vastus lateralis obliques (VMO and VLO) and RF during ramped isometric contractions (0, 50, and 70% MVC). They found that the muscle slide with respect to the skin showed an approximately 10 mm shift for the VMO and VLO. However, no study has investigated the effect of muscle contraction on the position of adductor muscles or how this could affect EMG electrode positioning relative to the muscle position.

Physiological crosstalk

It is important to recognise that bipolar surface EMG is not always a true representation of the electrical activity of a single muscle directly underlying the recording electrodes. With smaller or close muscles, the electrodes may pick up the electrical activity of one or more neighbouring muscles and their signals may become contaminated (known as crosstalk) with the surface EMG from the desired muscle (Figure 2- 6) (Winter et al., 1994). For example, the adductor muscles are close together, so crosstalk poses a considerable problem for this muscle group. While signal sources close to the electrode will dominate the recorded surface EMG signal, distant sources from other muscles may create crosstalk (Gerdle, Karlsson, Day, & Djupsjöbacka, 1999). The distance for effective EMG measurement is the radius about the electrode where the amplitude of signal contributions is larger than the standard deviation of the signal noise (Gerdle et al., 1999). The amplitude of the bipolar surface EMG signal

decays exponentially with increased distance from the recording electrode (Day, 1997). This

is due to the fact that muscle fibres, subcutaneous fat, and the skin are anisotropic and act as a

spatial filter with low pass frequency properties, where an increase in the distance between

the muscle fibre and electrode increases the filtering effect (Solomonow et al., 1994).

Effectively, this means that fewer signals can be measured from progressively more distant

electrical sources; consequently, the frequency of surface EMG contributions becomes

progressively lower (Lindstrom & Magnusson, 1977).
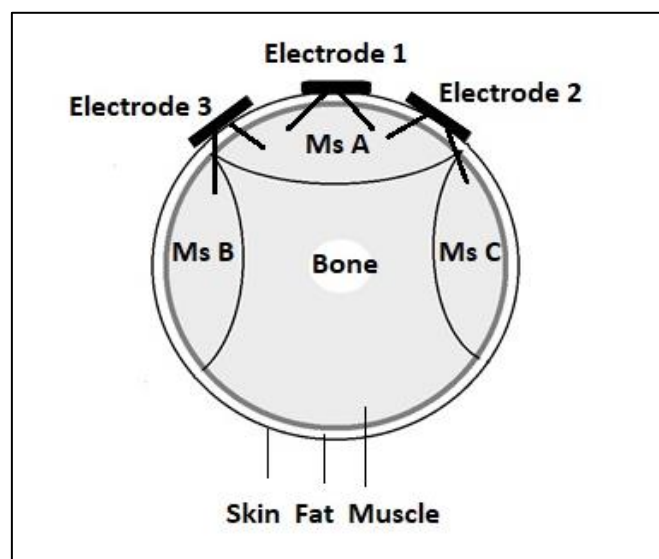
Figure 2- 6: Muscle A (Ms A), muscle B (Ms B), and muscle C (Ms C) are shown here. There are 3 surface EMG electrodes placed over the skin surface. Electrode 1 will pick up the best signal from Muscle A, while Electrode 2 and Electrode 3 would pick up cross-talk from adjacent muscles.

Crosstalk can be evaded by choosing proper-sized electrodes in terms of conductive area and

appropriate inter-electrode distance (Mesin, Merletti, & Rainoldi, 2009). Decreasing the

conductive area decreases the effective surface EMG measurement distance (i.e. depth).

Similarly, reducing the inter-electrode distance decreases the effective recording distance and

shifts the EMG bandwidth to higher frequencies (Lindstrom & Magnusson, 1977). The

SENIAM guidelines suggested a procedure for electrode placement for different muscles and

muscle areas (for more information see section 2.3 Experimental studies used to inform EMG

measurement protocols).

2.2.4 EMG acquisition

Electrode properties

The recording of EMG signals depends greatly on the properties of electrodes. In general, the electrodes have two main types monopolar or bipolar. In a monopolar arrangement, the sensors are placed on the muscle belly and the electrical reference point is grounded. This type of electrode is more susceptible to movement artefacts (Robertson, 2014). In contrast, the bipolar method utilises three electrodes: two electrodes are placed over the muscle and the third one is placed on the bony prominence. The electrical difference between the two electrodes is amplified (Robertson, 2014). In any type of application, electrodes can be surface or fine-wire electrodes (Figure 2-7) (Basmajian & De Luca, 1985). Surface electrodes are commonly used in kinesiological studies. They provide an overall representation of muscle activity as they record the activity for many muscle fibres underlying the placement area (Soderberg & Knutson, 2000). Apart from the benefit of easy handling, the main limitation of a surface electrode is that only surface muscles can be targeted (Aagaard et al., 2001; Jacobson, Gabel, & Brand, 1995). In contrast, fine wire electrodes are commonly used for deep muscles or small muscles where crosstalk is a potential problem (Jacobson et al., 1995).

Figure 2-7: Types of EMG electrodes: a) surface electrode, b) fine-wire electrode.

There is a special type of surface electrode known as array system electrode. It is a non-invasive system and considered a promising tool for characterising muscle properties (Kallenberg, Preece, Nester, & Hermans, 2009). Surface array EMG is recorded by placing a 1 or 2-D electrode array with closely spaced electrodes (<10 mm inter-electrode distance) on the skin overlying a muscle (Figure 2-8). With this type of application, information about motor unit (MU) anatomy (e.g. location of the innervation zone), MU size, and physiology (e.g. muscle fibre conduction velocity, recruitment strategy) can be obtained noninvasively (Drost, Stegeman, van Engelen, & Zwarts, 2006; Merletti, Farina, & Gazzoni, 2003; Thusneyapan & Zahalak, 1989).

Inserted/indwelling electrodes are fine-wire and needle sensors requiring invasive procedures; therefore, they are not commonly used to record EMG amplitudes for the superficial muscles during walking and running activities. A fine-wire electrode comprises of two small isolated wires with bared tips. They are offered in two forms: single wire or two wires. These wires are passed through a hypodermic needle for insertion with back-bent tips to form a barb to hold the sensor in the muscle when the needle is withdrawn. The distance between the bared tips sets the detection volume. Indeed, this type of electrode is rarely used to measure the activity of the adductor muscles because it is more susceptible to movement artefacts during dynamic tasks. In addition, signals recorded from a small area may not reflect the activity of the whole muscle (Bogey, Perry, Bontrager, & Gronley, 2000). Consequently, the actual role of a muscle during the examined task cannot be fully recognised. Therefore, the work in this project will use surface EMG electrodes.
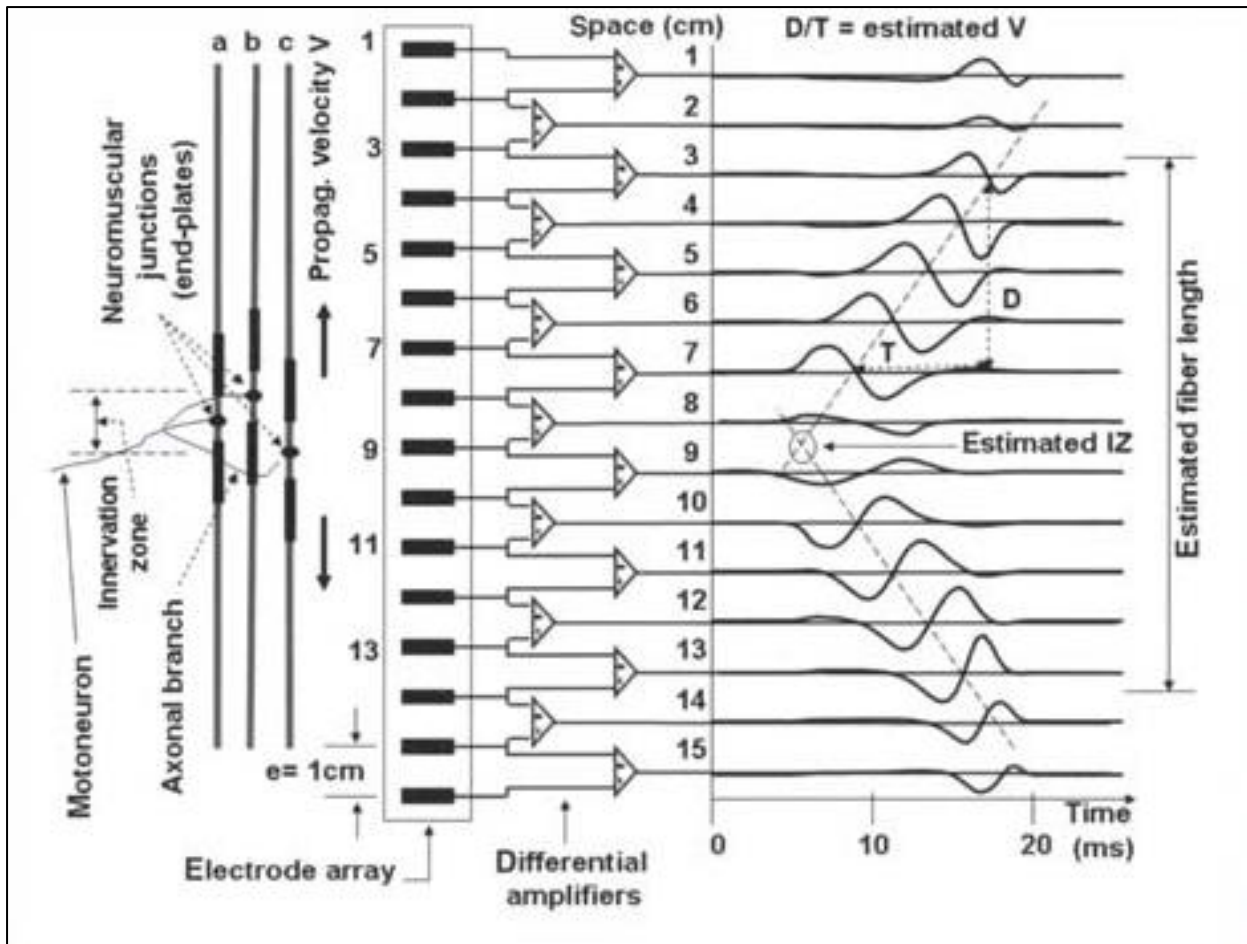
Figure 2-8: Example of a computer-generated motor unit action potential (MUAP) produced by a motor unit of three fibres (a, b, c). The MUAP is detected in differential mode from an array of 16 electrodes with an inter-electrode distance of 1 cm. Information on the innervation zone, fibre length, and conduction velocity of the MUAP can be obtained from the 15 signals.

2.2.5 EMG signal processing

The amplitude and frequency of the EMG signals are frequently investigated in a number of studies. Muscle activation occurs as a result of a number of active motor units and the frequency of activations. Therefore, a higher activation level will lead to an EMG signal with higher amplitude (number of motor units) and more frequent components (frequency of activation). Conventional EMG analysis, through the signal processing described below, combines these two features when determining the magnitude of the linear envelope.

Amplitude characteristics

The amplitude of EMG signals can be described by identifying the following five major variables. Firstly, peak-to-peak amplitude is suitable when the signal is highly synchronous, i.e. including multiple simultaneously firing motor units (Robertson, 2014). Secondly, the average rectified amplitude is the average of the absolute alternating current as the EMG signal occurs as an interference pattern with a zero average. Thirdly, the root mean square amplitude calculates the square values; therefore, not need for signal rectification. Fourthly, the linear envelope represents an approximation of the volume of activity. The EMG signal is a full wave rectified before passing through a low-pass filter. Cut-off frequencies between 3 Hz and 50 Hz have been proposed (Robertson, 2014). Additionally, 10 Hz displays a satisfactory waveform for short-duration activity (Robertson, 2014). As high-frequency waves are attenuated by the signal, the residual signal may be improper for the analysis of the onset-offset of the profile. Finally, integrated EMG is the summation of the accumulated activity over a period of time.

Frequency characteristics

The frequencies in EMG signal can be illustrated by turning points and zero crossing. The turning points method sums the number of peaks per unit of time in addition to the repetition in which the signal crosses the zero level. The repetition is interrelated with other frequency variables such as spectral analysis (Inbar, Paiss, Allin, & Kranz, 1986). The mean and median frequency or spectral analysis techniques are frequently used. A positive skewness with an estimated mean and median of 120 Hz and 100 Hz, respectively, is commonly reported in surface EMG measurements (Robertson, 2014). Alterations to the previous criteria can be used to specify the variations in the conduction property of the muscle fibres. Another frequently used technique is the onset-offset analysis. The onset-offset method locates the starting and ending points of muscle activation (Strazza et al., 2017; Sutherland, 2001). For

this purpose, the EMG signal should not be filtered or processed, as this may reduce high-frequency waves. Moreover, several methods have been used to analyse the EMG signal for specific purposes such as recurrence quantification analysis (Filligoi & Felici, 1999), neural network classification (Liu, Herzog, & Savelberg, 1999), and wavelet analysis (Karlsson, Yu, & Akay, 1999).

2.2.6 Limitations of EMG measurement

Although EMG affords valuable information on the neural control of human movement, it has several limitations. A major limitation is that it does not give information about muscle force (Hug, Hodges, & Tucker, 2015). As the EMG method is the study of electrical and not mechanical activity, it does not present a direct measure of the force or torque being generated by the tested muscle (Winter, 2009). The physiological criteria of the human muscle can also affect the interpretation of EMG signals. For instance, at the same activation level, a given muscle can generate more force depending on its PCSA, optimal length, and the velocity of contraction (Hug et al., 2015). Another limitation associated with EMG recordings is the electromechanical delay between the muscle activation and the subsequent force production. The assumption of instantaneous mechanical effect of the muscle when a muscle is activated misleads the interpretation of EMG data. This delay must, therefore, be considered when attempting to identify muscle activation characteristics that support against external loads during dynamic tasks (Hug et al., 2015).

Despite the limitations of EMG measurements, EMG is still an effective electrodiagnostic tool for characterising muscle activation. Two types of EMG electrodes are commonly used in order to detect the signal from muscles: surface and fine-wire electrodes. The surface electrode is the best electrode to record the activity of the adductor muscles as the application of fine-wire electrodes between the legs during gait is difficult. In addition, there is a lack of

guidelines for measuring the adductor muscles using surface electrodes, so there is a need to set a standard method for this purpose. Moreover, many issues in surface EMG measurement such as physiological crosstalk, soft tissue signal attenuation, and muscle movement relative to the skin can complicate the measurement of the EMG activity of the adductors.

2.3 Experimental studies used to inform EMG measurement protocols

Acquisition of EMG signals in dynamic conditions is influenced by physiological and non-physiological factors that are common to all EMG data acquisition processes (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). These factors include subcutaneous tissue layers, muscle fibre conduction velocity, spread of the innervation zones (IZ) among motor units, crosstalk from nearby muscles, electrode size and shape, inter-electrode distance, and the location of the electrodes along the muscle (Farina et al., 2004). The latter affects the consistency of EMG measurements and interferes significantly in the statistical and spectral characteristics of the EMG. The failure to adhere to optimal electrode placement will result in erroneous interpretations of these signals (Farina et al., 2004; Farina, Merletti, Nazzaro, & Caruso, 2001; Merletti, Rainoldi, & Farina, 2001; Roy, De Luca, & Schneider, 1986).

Given the potential difficulty with EMG measurement and the importance of positioning EMG electrodes in the correct position, a range of experimental techniques have been proposed to inform protocols on EMG data collection. Such techniques originated with the classical recommendation of electrode placement (Basmajian & De Luca, 1985). Thereafter, a project on sensor location, SENIAM guidelines, was developed. This was followed by a number of studies aimed at monitoring the shift of the IZ sites during dynamic conditions. These techniques will be discussed in the following paragraphs.

Since electrode placement determines the electrical view of a muscle, consistent placement of the electrodes is important for EMG measurements. In 2000, European concerted action in the Biomedical Health and Research Program of the European Union developed a project for sensor location called a SENIAM. This project aimed to guarantee the repeatability of EMG assessments and interpretations. The SENIAM recommendations for 22 different muscles were based on the results of an inventory, a workshop, and experimental studies performed by SENIAM's members (Hermens et al., 1999). Sensor location referred to the position of the two bipolar sites overlying a muscle in relation to the line connecting two anatomical bony prominence (Hermens et al., 2000).

The goal of sensor placement is to place the electrode in a good and stable location where the high-quality EMG signals can be gained. The location of sensors is based on the following principle: with respect to the longitudinal (in fibre direction) location of the sensor on the muscle, it is proposed to centrally place the sensor between the (most) distal motor endplate zone and the distal tendon. Additionally, in the term of the transversal location of the sensor on the muscle, it is suggested to place the sensor away from the muscle border so that the geometrical distance to other muscles is maximised (Day; Merletti, Hermens, & Kadefors, 2001; Stegeman & Hermens, 2007). However, the SENIAM specifies the best locations for electrode placement for most of the lower limb muscles, but it did not take into account the IZ shift especially during the dynamic task. The variations in muscle length and joint angles during the ambulatory task will lead to a shift of the IZ, resulting in poor-quality EMG signals (Campanini et al., 2007; Farina et al., 2001; Nishihara, Kawai, Chiba, Kanemura, & Gomi, 2013).

The linear-array electrode has been recommended as a technique to recognise the IZ locations along the muscle (Figure 2-9). This is achieved through several surface EMG signals

simultaneously propagated along a line (Figure 2-8) (Farina, Fortunato, & Merletti, 2000; Merletti et al., 2003; Nishihara, Kawai, Gomi, Terajima, & Chiba, 2008). With the linear array, surface EMG is recorded by placing a 1 or 2-D electrode array with closely spaced electrodes; this method is used to obtain high degree consistency of the EMG variables for muscle characterisation (Kallenberg et al., 2009; Rainoldi, Bullock-Saxton, Cavarretta, & Hogan, 2001; Rainoldi et al., 1999). However, with the linear-array electrodes, it can be hard to maintain constant interaction for the large number (64 to 121) of electrodes either due to bad electrode–skin contact or short circuit between two or more surface electrodes (Marateb et al., 2011). In addition, the comparatively large size electrode makes them difficult to use with small muscles. Therefore, the electrode most commonly used in the literature and recommended by the SENIAM is still the bipolar electrode. Consequently, developing a technique of bipolar electrode placement confirming its precise location during dynamic contractions (Rainoldi, Melchiorri, & Caruso, 2004) would be a contribution that is valuable with widespread benefit.
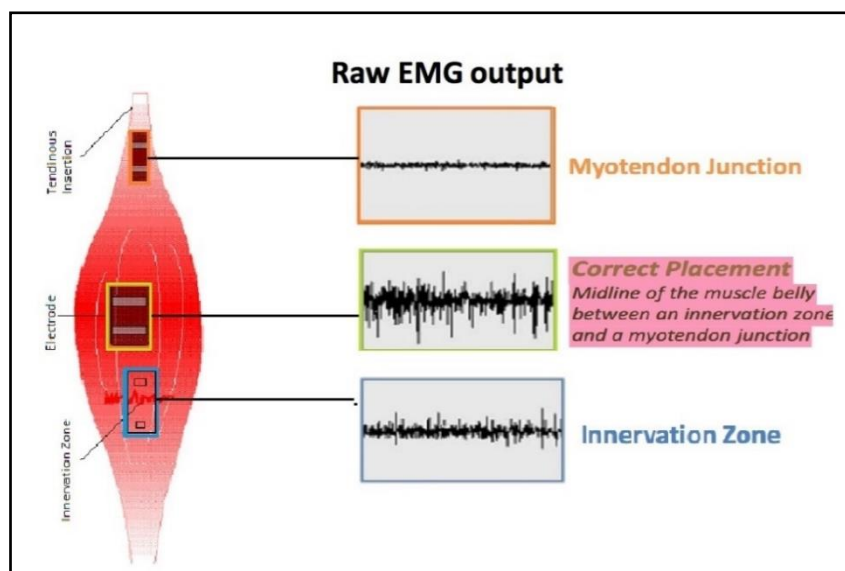
Figure 2-9. Best electrode location over the skeletal muscle.

In light of the diverse positioning of the electrode and importance of exact EMG electrode placement, many researchers used ultrasound imaging in order to determine the best locations

for surface and needle electrodes in humans. For decades, ultrasound imaging has been used to describe the structural and morphological changes in skeletal muscles. Recently, ultrasound imaging was used to measure subcutaneous adipose tissue covering the muscles (Wu, Delahunt, Ditroilo, Lowery, & De Vito, 2017). In addition, it is used as a tool to increase the accuracy and verify correct placement of needle electrodes (Corneil, Goonetilleke, Peel, Green, & Welch, 2012). For this technique, a standard ultrasound device with a linear probe is used to visualise the muscles. In this way, feedback about electrode placement is acquired during the produce, allowing correct placement of the needle in the selected muscle.

A similar technique was introduced to locate the placement area for skin EMG electrodes, especially for the muscles that were not covered in the SENIAM guidelines. For example, Watanabe et al. (2009) used ultrasonography to define the structural properties of the superficial area of the hip adductor muscles to finally place the surface EMG electrode. Using the same technique, four hip adductor muscles (AL, AM, pectineus and Gr) were located via real-time ultrasonography in order to measure their EMG activity during six different examination tests (Lovell, Blanch, & Barnes, 2012). Using the ultrasound technique for detecting electrode placement, particularly for the adductor muscles, is a possible solution for a number of measurement problems that arise during dynamic tasks. The first problem is that there is no clear guideline on surface EMG electrode placement for the adductor muscles. Moreover, the individual hip adductor muscles are adjacent to each other at the upper medial aspect of the thigh (Watanabe et al., 2009). Thus, small errors in electrode placement could increase the possibility of crosstalk, defined as the picking up of signals from a nearby muscle rather than the muscle over which the electrode is placed. Another option is to use fine-wire electrodes which are inserted directly into the muscle under the direction of

ultrasound (Mann et al., 1986; Montgomery et al., 1994). However, this invasive technique may not be appropriate for the adductor muscles, especially during dynamic tasks.

It is currently not clear how much movement of the adductor muscles occurs relative to the overlying skin during movements typical of gait. As the adductor muscles are closely located in an area on the inner side of the thigh, there is a high likelihood of picking EMG signals from the neighbouring muscles when using surface electrodes. By quantifying the adductor movement in different hip flexion/extension angles, it would be possible to understand the movement of the adductor muscles relative to the skin and therefore make an informed judgement on the possibility of crosstalk. Therefore, the main aim of the first part of this thesis was to obtain a standardised method for measuring this group of muscles in different lower limb positions, typical of walking and running. Three separate but complementary experiments were conducted in order to quantify adductor muscle movement relative to the skin to determine optimal EMG electrode placement.

2.4 Force-EMG relationship

The previous literature has shown that there is a relationship between the torque produced by muscle contraction and its corresponding EMG amplitudes (Billot, Simoneau, Van Hoecke, & Martin, 2010; Maganaris, Baltzopoulos, & Sargeant, 1999). In the skeletal muscles, this relation is based on motor unit recruitment, motor unit firing rate, crosstalk, differences in the location of the recording electrodes, and the participation of synergistic muscles in force generation (Kuriki et al., 2012; Solomonow et al., 1990). Under isometric contraction, incremental changes in muscle forces are linked with the changes in EMG amplitudes (Bouisset & Maton, 1972; Jacobs & van Ingen Schenau, 1992; Milner-Brown & Stein, 1975). In dynamic contractions, the relationship between EMG and force has a greater complexity due to experimental condition and the physiological characteristics of the tested muscles.

Nevertheless, dynamic investigation between force and EMG amplitude is very challenging and so it is first necessary to studies force-EMG relationships under dynamic conditions.

The suitability of the new locations proposed to locate the adductor muscles in order to accurately place the surface EMG electrodes needs to be verified. Obtaining relation during a ramped isometric contraction is one of the methods to validate the EMG measurement. De Luca (1997) suggested that if the newly recruited motor unit is situated far away from the electrode, then the force will increase, but the amplitude of the EMG signal will not. This will result in poor or no correlation between the EMG amplitude and the force produced by the corresponding muscles. Therefore, the third experiment of the first study aimed at investigating the relationship between the torque produced by the adductor muscles and the magnitude of the corresponding EMG signal during isometric contraction.

2.5 Normalisation and analysis of the EMG signal

Normalisation is a critical step in processing the EMG signals. This process, the normalisation, allows the comparison of EMG signals of a certain muscle on different testing days or among different individuals. There are several ways to normalise the EMG signals. The following section focusses on exploring the different methods of normalisation. In addition, it highlights the advantages and disadvantages of each method and the most appropriate method that should be applied for the adductor muscles.

2.5.1 Definition and importance of normalisation

Normalisation of EMG signals is a procedure by which the electrical signal values of activity are expressed as a percentage of that muscle's activity during a calibrated test contraction (Lehman & McGill, 1999). Normalisation is executed by dividing the EMG amplitudes gained from a specific task or event by the EMG amplitudes of a reference contraction of the

same muscle and this will be presented as a proportion or percentage (Burden, Trew, & Baltzopoulos, 2003). The process of normalisation is needed in order to enable a comparison among electrode sites on the same muscle or on two different muscles and for documenting variation over testing days. In fact, normalisation is necessary for any comparative analysis of EMG amplitudes (Mathiassen, Winkel, & Hägg, 1995). This thesis aimed to establish a new method for measuring the activity of the adductor muscle group and data from different subjects were collected. It was therefore necessary to identify the most appropriate method for normalising the EMG signals collected from this muscle group.

The basic idea of normalisation is to calibrate the microvolts value to a unique calibration unit with physiological relevance. For individuals with normal neuromuscular control, one of the most convenient references of normalisation is the EMG generated during the maximum voluntary effort. The outcome is then reported as percentage of maximum voluntary contraction (%MVC). The application of normalisation was undertaken to account for variability among recording factors (e.g. changes in skin-electrode impedance, subcutaneous tissue thickness, and variations in electrode placement) and to facilitate comparisons between individuals and muscles. It is preferable to use this technique in order to compare the effect of a certain intervention on the EMG activation pattern (Lehman & McGill, 1999). The outcome of normalisation approaches is that the influence of the given detection conditions is eliminated and data are rescaled from microvolt to percentage of the selected reference value. It is imperative to understand that amplitude normalisation does not change the shape of EMG curves, but only their Y-axis scaling.

2.5.2 Different methods of normalisation

As mentioned earlier, because of the natural variability of EMG signals, normalising the EMG signals is essential for physiological interpretation and comparison between muscles

and between individuals. Previous studies used a wide range of approaches to produce reference EMG values for normalisation purposes that can be duplicated across subjects and different testing days, including isometric, isokinetic, and dynamic muscle actions (Figure 2-10).

a) Isometric testing

Maximum and submaximum isometric contraction

Maximal isometric voluntary contraction (MVIC) is the normalisation method suggested by the SENIAM and Kinesiology's guidelines. It is the most broadly employed normalisation technique during which EMG amplitude is expressed as a percentage of the maximum neural activation of the desired muscle (Burden et al., 2003; De Luca, 1997). This strategy is considered a powerful approach for physiological interpretation of signals in a healthy population. However, generating the MVIC is not always possible, e.g. for older people or for patients with musculoskeletal disorders. Additionally, achieving the maximum force output does not mean acquisition of a maximal electrical activity for a given muscle (Lehman & McGill, 1999). Furthermore, the majority of studies have displayed poor EMG reproducibility both within and between participants and between days with this type of normalisation (Ball & Scurr, 2010; Bamman, Ingram, Caruso, & Greenisen, 1997; Heinonen et al., 1994; Yang & Winter, 1984). This could occur as a result of the onset of fatigue, synergistic contribution, and psychological factors (Enoka & Fuglevand, 1993; Miaki, Someya, & Tachino, 1999).

The use of the MVIC as a method of normalisation is associated with several technical concerns. Such concerns could have an influence on the validity and repeatability of the normalisation procedure. These concerns are the inertial effects at the onset of the test, patient fatigue, position of subject during testing, and the kind of motivation used. Normalisation is

not a measure of muscular force, but it is a measure of muscular activation level relative to the maximum activity of the tested muscle obtained during MVIC (Soderberg, 1992). Therefore, the normalised EMG data obtained from the MVIC cannot reflect the maximum activation capacity of the muscle at lengths differ from those at which the MVIC was recorded or under non-isometric conditions. Accordingly, De Luca (1997) recommended the use of less than 80% of the MVIC in order to obtain normalised EMG amplitudes aiming to provide a suitable reference point. However, the study did not provide a sufficient detailed protocol on how the submaximal MVIC was calculated. Using sub-maximal load results in improvement of between day reliability as compared to when using maximal load for knee extensors and triceps (Rainoldi et al., 1999; Yang & Winter, 1984). However, MVIC is still the best method if an overall measure of the level of activation is required.
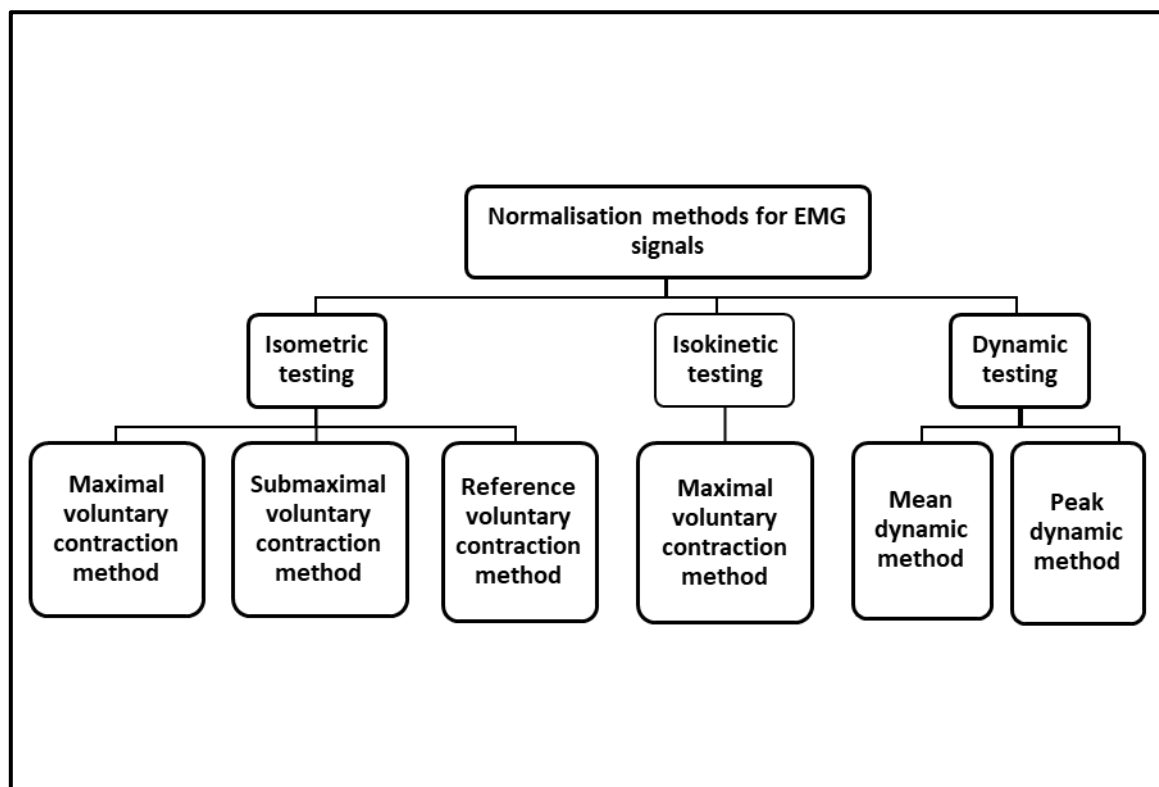
Figure 2-10: Diagrammatic scheme represents the different EMG normalisation methods.

Reference voluntary contractions method (RVC)

This technique is used for the clinical subjects who are incapable of performing maximal level of muscle contractions or who need an analogous controlled task for interpreting repeated tests. During this normalisation process, the participant assumes a specific posture and performs a specific task. All these procedures are repeated at every testing session. For instance, assuming a standing posture, the participant was instructed to hold up a precise weight (0.5 kg) in each hand with arms flexed (upper arms parallel to the floor, lower arms perpendicular) for 10 sec (Figure 2-11) (Lewis, Holmes, Woby, Hindle, & Fowler, 2012). With this approach, the external load is controlled and this is likely to minimise variations in myoelectric activity between testing days. While changes in raw EMG amplitude (non-normalised) could results from any of the modulators and artefacts noted in the previous section, changes in the normalised EMG amplitude in the RVC indicate a true variation in the neural control pattern (Lehman & McGill, 1999).

Figure 2-11: A subject performing a reference voluntary contraction (Lewis et al., 2012).

The main restraint of using the submaximal isometric contractions and the RVC is that comparing the activity levels between muscles and among individuals are not valid because the reference value measured in these methods is not relative to the maximum capacity of the muscle. For example, lifting an absolute weight such as 2 kg might necessitate a 20% of the maximum muscle capacity in a strong individual while in another subject who is not as fit may need a 50% of his maximum muscle capacity to lift the same load. Another limitation is that the neuromuscular control strategy may be vary between subjects or between sides within the same subject (Ounpuu & Winter, 1989). This factor does exist during the maximal contractions where all possible fibres are engaged to achieve maximum force generation. Therefore, these methods cannot improve the reliability of EMG measurements, as they do not allow for accurate comparisons between muscles or individuals.

b) Isokinetic testing

EMG signals from isokinetic muscle testing have been suggested as an alternative normalisation technique for EMG amplitudes. In this technique, an individual performs a maximum isokinetic contraction at a speed similar to the dynamic task under investigation (Mirka, 1991). The activation level versus the joint angle curve generated from the maximum dynamic contraction is then used to normalise the EMG data. This procedure allows the quantification of the joint angle, torques, and corresponding EMG amplitudes (Kellis & Baltzopoulos, 1996; Mirka, 1991). A good EMG reliability has been reported between trials with isokinetic exercises for the knee extensors and flexors (Finucane, Rafeei, Kues, Lamb, & Mayhew, 1998; Larsson, Karlsson, Eriksson, & Gerdle, 2003). In contrast, Ball and Scurr (2010) found a poor reliability between days and weeks of isokinetic testing of the triceps surae muscles. The results obtained using this technique for normalisation are less reliable than those obtained from other methods (Burden et al., 2003).

c) Dynamic testing

Peak dynamic and mean dynamic methods

These methods are commonly used in the investigation of dynamic tasks such as walking and running. For the peak dynamic technique, each point that constitutes the processed EMG is divided by the peak value acquired from the same EMG (Arendt-Nielsen, Graven-Nielsen, Svarrer, & Svensson, 1996; Van Hedel, Tomatis, & Müller, 2006). Alternatively, Albertus-Kajee, Tucker, Derman, Lamberts, and Lambert (2011) calculated the peak amplitude obtained from a number of separate testing trials of a certain dynamic task performed with a maximum speed (e.g. sprint). This procedure was used as an alternative method for normalising the EMG amplitudes of the same dynamic task performed with a specific speed (e.g. running at 70% of the maximum speed). During this method, subjects performed two maximal 20m sprints. Then, the EMG from the fastest sprint was analysed by isolating three peak amplitude contractions from the middle of the recorded sprint. The resultant amplitudes were averaged and used for normalisation. This alternative method results in decreased inter-individual variability.

Similarly, Yang and Winter (1984) used a number of normalisation methods. The standard for choosing the best normalisation technique was highest reduction of the inter-subject variability of the EMG signal. The researchers pointed out that the peak and mean dynamic methods could reduce the effect of individual-specific and situation-specific conditions, which may give rise to signal variance. In addition, Ball and Scurr (2010) found that use of these normalisation methods provided reliable EMG amplitudes both between days and between weeks for the triceps surae muscles during sprint running. Normalising to the peak or mean amplitude during the activity of interest has been shown to reduce the variability between subjects relative to using raw EMG data or using amplitudes normalised to MVIC

(Allison, Marshall, & Singer, 1993; Chapman, Vicenzino, Blanch, Knox, & Hodges, 2010; Winter & Yack, 1987; Yang & Winter, 1984). However, the decrease in the variability between subjects by normalising to the peak or mean amplitude recorded during an activity is achieved by eliminating some biological variations (e.g. strength difference) between subjects (Allison et al., 1993; Knutson, Soderberg, Ballantyne, & Clarke, 1994). In addition, normalisation to peak or mean of the dynamic activity could remove information on the overall level of activation and therefore may hide actual differences between subjects.

2.6 Repeatability of lower limb EMG measurement during walking and running

This thesis aimed to develop a new method for locating the adductor muscles in order to record their EMG activity. As such, there was a need to quantify the repeatability of this method. Therefore, the following section focuses on the issue of EMG repeatability and its importance. Moreover, the relevant studies discussing the EMG repeatability during human ambulation, especially during walking and running tasks will be outlined. Finally, the factors that influence the repeatability of EMG measures will be identified.

2.6.1 Definition and importance of repeatability

Repeatability is an imperative consideration when using gait analysis data as an adjunct to clinical decision-making. A repeatability study is conducted to assess the degree to which measurements vary when they are repeated on the same subject. Previous studies have shown that the reliability of measurements is better within a test session than between different test days (Kadaba, Wootten, Gainey, & Cochran, 1985; Kadaba et al., 1989). The reduced consistency has been attributed to the difficulty of replacing the electrode on different days, despite the exactness of the investigator (Kadaba et al., 1985). The study of muscle electrical activity and its reproducibility permits researchers to understand the neuromuscular

mechanisms controlling normal gait and to establish the normal EMG profile for use as a reference pattern in pathological conditions. For clinical applications, the repeatability of EMG signals is essential to evaluate and to categorise neuromuscular deficits. Therefore, assessment of the reliability of EMG variables is of considerable relevance to the clinical and experimental use of such a technique (Laplaud, Hug, & Grélot, 2006).

2.6.2 Repeatability of EMG measurement during walking

A limited number of studies have determined the repeatability of EMG measurement during walking tasks. Kadaba et al. (1989) found that repeatability within one particular day was slightly better than between test days for the GMax, GMed, AL, VLO, RF, VMO, lateral hamstrings (LHam), medial hamstrings (MHam), tibialis anterior (TA), and medial gastrocnemius (MGastro) muscles. The EMG amplitudes were normalised with respect to the maximum amplitude within each gait cycle. They calculated the Coefficient of Multiple Correlation (CMC) and found a high within-day and between-days reliability (>0.8, 0.7 respectively) for all muscles, although this was lower for the AL (0.7, 0.6 respectively). Lyytinen et al. (2016) demonstrated that the VMO, TA, and biceps femoris exhibited a good between-day repeatability (ICC ranged from 0.77 to 0.84) in level walking when EMG was normalised to the activation estimate of the maximum EMG signal obtained during walking upstairs at 0.5 m.s^{-1}. Murley, Menz, Landorf, and Bird (2010) concluded that the normalisation techniques have an impact on the within-session reliability of EMG parameters for the leg lower muscle muscles. The TA and MGastro displayed good to excellent reliability when normalised to sub-maximal voluntary contraction compared to normalisation to maximum voluntary contraction. The TA displayed moderate reliability when normalised to sub-maximal voluntary contraction (ICC: 0.34–0.56) compared to good to moderate for MVIC normalised values (ICC: 0.56–0.65). The MGastro displayed good to very good

reliability for MVIC normalised values (ICC: 0.61–0.84) and poor reliability for sub-maximum values (ICC: 0.08–0.19).

The reliability of EMG characteristics during walking has also been determined in patients with musculoskeletal conditions. Hubley-Kozey, Robbins, Rutherford, and Stanish (2013) found good to excellent ICC values (> 0.8) in patients with moderate knee OA for the lateral gastrocnemius (LGastro) and MGastro, VMO, VLO, RF, and MHam and LHam muscles. Similarly, the EMG profile of the MGastro exhibited excellent between-days repeatability during level walking for OA individuals (Lyytinen et al., 2016). In addition, the CMC values were higher in the shod condition than the barefoot condition in rheumatoid arthritis patients associated with pes planovalgus for TA, soleus, peroneus longus, and MGastro muscles (Barn, Rafferty, Turner, & Woodburn, 2012). However, no study has examined the reliability of EMG measurement for all three adductor muscles during walking for either healthy participants or people with musculoskeletal disorders.

2.6.3 Repeatability of EMG measurement during running

Similar to studies investigating the reliability of EMG measurement during walking, only a small number of studies have been carried out to detect the reliability of EMG measurement during running. These studies aimed to determine the reliability of EMG parameters during running within a single session. For example, Smoliga, Myers, Redfern, and Lephart (2010) recorded EMG from the legs (VLO, semimembranosus, GMax, and RF), torso, and arm muscles during running. They found good reliability (ICC > 0.80) for the parameters studied which included integrated EMG, root mean square EMG, maximum M-wave (defined as the maximum magnitude of the absolute value of the band-pass filtered time-domain EMG signal), and median power frequency. EMG repeatability has also been assessed during treadmill running at different velocities for five lower-extremity muscles (Karamanidis,

Arampatzis, & Bruggemann, 2004). These studies found that the reproducibility of the EMG data for the posterior leg muscles is better than for the anterior leg and thigh muscles (i.e. the MGastro and LGastro showed high reliability, while the VLO, hamstrings, and TA demonstrated low reliability).

In the same context, Golhofer, Horstmann, Schmidtbleicher, and Schoenthal (1990) showed good intra-individual stability for the gastrocnemius and soleus during running. The activity of the VMO, VLO, RF, biceps femoris, and MGastro appears to be more repeatable when normalised to maximum voluntary contraction or peak running speed (ICC > 0.80), compared to normalisation to 70% peak running speed (Albertus-Kajee et al., 2011). The between-day repeatability of the EMG profile for the adductor muscles during running has not been discussed in the literature. Therefore, there is a need to understand the standard level of EMG repeatability for this group of muscles.

2.6.4 Factors affecting the repeatability of EMG measurement

The repeatability of EMG measurement during dynamic situations is influenced by many factors. These factors include body posture during the performed task, the method of normalisation, session-based testing, type of electrode used, familiarisation effect, and visual feedback (Dankaerts, O'Sullivan, Burnett, Straker, & Danneels, 2004). In terms of the method of normalisation, it has been found that sub-maximal voluntary isometric contractions (SVICs) are more reliable than maximum voluntary isometric contraction (MVICs) (Dankaerts et al., 2004; Ha, Cynn, Kwon, Park, & Kim, 2013; Kollmitzer, Ebenbichler, & Kopf, 1999). Additionally, the test-retest reliability for both MVICs and SVICs has high intra-session but only SVIC test-retest reliability was high inter-session (Dankaerts et al., 2004). For session-based testing, it appears that inter-session reliability is much worse than intra-session test-retest reliability (Dankaerts et al., 2004; Kollmitzer et al., 1999; Oskouei,

Paulin, & Carman, 2013). Moreover, surface electrodes tend to display slightly better reliability than fine wire electrodes (Burnett, Green, Netto, & Rodrigues, 2007). It has been proposed that a lack of familiarity may impede the test-retest reliability of MVICs (Ball & Scurr, 2013), while using a visual feedback for the level of activation being produced markedly increases the reliability of the SVIC condition (Burnett et al., 2007). In order to ensure that subjects reached the required level of the SVIC, a computer monitor, as a biofeedback method, was positioned in the subject's field of view in order to provide appropriate feedback. The SVIC level then was displayed on the computer screen as a solid line and each subject was asked to produce torque matched with his pre-determined SVIC level (Burnett et al., 2007). This technique improves the reproducibility of the SVIC and consequently enhances the reliability of the EMG signals.

In summary, the reliability of surface EMG during walking and running has been established for healthy subjects. However, the previous studies were restricted to specific lower limb muscles and the adductor group was excluded from these studies. In addition, while the previous studies have demonstrated a good intra-session reliability, the reliability between different testing sessions is not explored. Accordingly, there is a need to demonstrate the acceptable standard for between-day EMG reliability for the adductor muscles and the major lower limb muscles during walking and running.

2.7 EMG profile for the adductor muscles

The previous sections focused on EMG measurement for the adductor muscles. In addition, the previous research attempted to address the difficulties encountered by the researcher when measuring the activity for this group of muscles. This section is an attempt to understand the normal function of the adductor muscles and how they behave during walking and running activities. This was achieved by reviewing the articles identified in the

systematic search strategy (see Chapter 1). Three articles for walking, four articles for

running, four articles for both walking and running, and eight articles for the musculoskeletal

disorders were identified (Figure 2-12). This number of articles is quite low compared to the

number published for the other lower limb muscles, such as the hamstring or quadriceps.

Walking
- 3 articles
- Number of participants: 18, 17, 11
- Type of electrode: fine wire (1) – Surface (2)
- Muscle measured : AM (1)- AL (3)- Gr(1)- AB(1)

Running
- 4 articles
- Number of participants: 20, 11, 9, 30
- Type of electrode: fine wire (1) – Surface (3)
- Muscle measured : AM (1)- AL (1)- Hip adductors without specifications (2)

Walking and Running
- 4 articles
- Number of participants: 34, 15, 10, 34
- Type of electrode: fine wire (1) – Surface (3)
- Muscle measured : AL (1)- Hip adductors without specifications (3)

Musculoskeletal disorders (MSDs)
- 8 articles
- Type of MSDs: Acetabular labral tear (ALT) (2)- Groin pain (1)- Hip pain (HP)(1)- Hip OA (1)- PFPS (3)
- Number of participants: (20 Healthy (H)-20 PFP), 16, (21 H-10 PFP), (17 ALT- 17 H), (22 PFP), 1, (9 HP, 9H), (16 Hip OA)
- Task performed: Stair ambulation (1)- Ex's (2), Lunge(1), ES (1)- Walking (1), Standing hip flexion(1). Walking with APOS (Novel biomechanical shoes) (1)
- Type of electrode: fine wire (2) – Surface (6)
- Muscle measured AM (2)- AL (6)

Figure 2-12: This diagram represents the walking, running, and musculoskeletal disorder articles identification by systematic search strategy for the adductor EMG activity. AL. adductor longus; AM. Adductor magnus; AB. Adductor brevis; Gr. Gracilis. Hip OA. Knee osteoarthritis; PFPS. Patellofemoral pain syndrome

Averaged EMG profiles in walking and running at different speeds

Only a small number of published studies describe typical EMG profiles for the hip adductor

muscles. Such profiles were measured using either surface or fine electrodes. Winter and

Yack (1987) proposed that during the stance phase of walking at, two adductor muscles (AL

and AM) have a moderate activity in order to stabilise the hip joint against the action of the

hip abductors. They reported peak activity of longus and magnus during the swing phase at a

self-selected speed and suggested this functioned to control the lateral movement of the

swinging limb (Figure 2-13). This peak occurs at approximately 70% and 80% of stride for

the AL and AM respectively. The AL may peak early in swing because of its additional role as a hip flexor (Winter & Yack, 1987). In contrast, Perry (2010) observed the activity of the AL and Gr to occur in late stance during free cadence walking (self-selected speed) and to persist throughout the swing phase and the AM to activate in early stance (Figure 2-14). Compared to AL, the onset of AM activity occurs in terminal swing with the peak activity at or just before initial contact. This shows that there is inconsistency and also that there is minimal information available on inter-subject variability. In addition, there is a minimal data describing a normal EMG profile for the adductor muscles, and there is no consensus on differences in adductor activation patterns across individuals. Therefore, more research is needed.
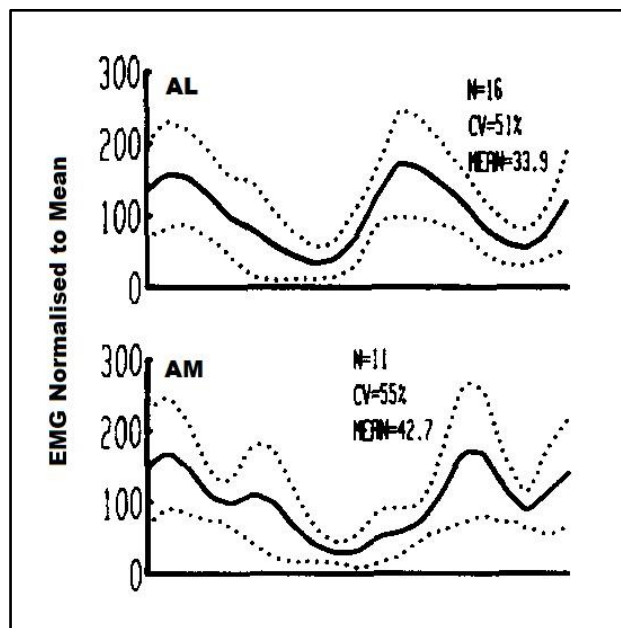
Figure 2-13: EMG profile for the adductor longus (AL) and adductor magnus (AM). Intensity is displayed as a percentage of the mean dynamic test value adopted from (Winter & Yack, 1987).

More strenuous activities, such as jogging, running, and other athletic events, place a greater functional demand on the adductors. Montgomery et al. (1994) proposed three peaks of activity for the AM during running at 4.2 m.s^{-1}; the highest peak occurring early in stance during the loading response, the second peak during early swing, and the third peak in middle

swing (Figure 2-15). Moreover, surface EMG of the AL in healthy individuals was studied

during different phases of run-to-cut manoeuvres in the stance leg (Chaudhari, Jamison,

McNally, Pan, & Schmitt, 2014). The results revealed that the AL activates just before the

initial contact and continues during the early and middle stance. The activity of the AL begins

again in late stance and continues in the early swing (Chaudhari et al., 2014). However,

electrode placement has been used to target the AL as a surrogate for all hip adductor

activity, due to the close proximity of the hip adductors and difficulty in pinpointing the

activation of individual adductor muscles with surface electrodes.
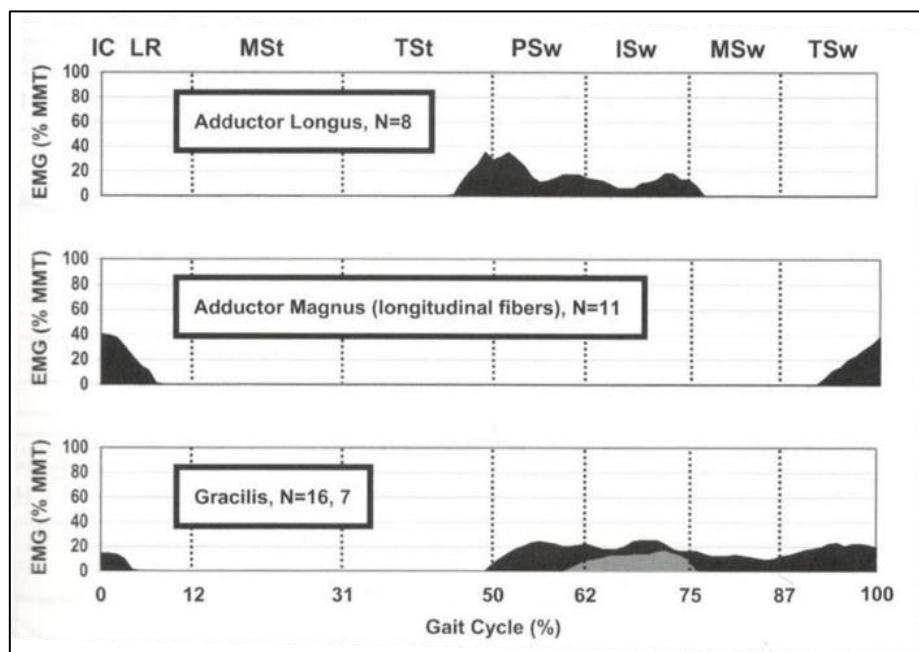
Figure 2-14: Hip flexor muscles. Normal mean intensity and timing during free walking (quantified electromyogram). Intensity as a percentage of maximum manual muscle test value (% MMT) indicated by the height of the shaded area. The dark shading indicates the activity of the majority of subjects. The light grey area indicates a less frequent activity pattern. Vertical bars designate the gait phase divisions. N= samples included in data (Perry, 2010).
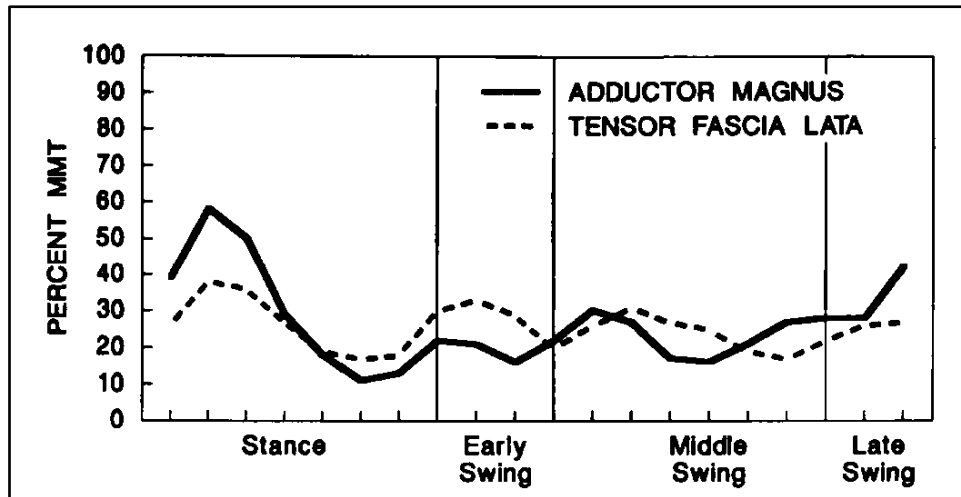
Figure 2-15: The EMG data for the adductor magnus and tensor fascia lata muscles during training pace.

A few studies have demonstrated the effect of gait speed on the adductor muscles. For example Mann et al. (1986) compared the activity of the AL muscle using fine wire electrodes during walking (1.32 m.s^{-1}), and running (4.77 m.s^{-1}) activities. They found that the period of activity of the AL began just after toe-off and continued into the early forward swing. In addition, AL demonstrated a greater period of activity during running. At all speeds, it became active during toe-off and remained active during follow-through and the early forward swing. Another study examined AM activity during walking and running (Gazendam & Hof, 2007). The AM shows a profile in which three peaks can be distinguished, in mid-stance at 18%, in mid-swing at 68%, and in final swing at 90%. These peaks become prominent only while running with a speed of 3 m.s^{-1} and higher; at lower running speeds the EMG is low and irregular, even to the extent that it is hard to see a periodicity. In walking (2.25 m.s^{-1}) the pattern is very different from running, with peaks at foot contact (0%) and toe-off (60%) (Gazendam & Hof, 2007).

The aforementioned studies reveal a high level of variability among researchers in their understanding of how the adductor muscles activate during walking and running activities. The reason for this variability may be the lack of standardisation in the method used to record

the activity of these muscles. It could also be due to the use of two different recording EMG electrodes (fine wire and surface electrodes). Therefore, it may be possible for gain a better understanding of how this group of muscles works during gait by having a standard method for recording their activation pattern using surface EMG electrodes. Therefore, the study described in the next section aimed to describe the EMG profile for the adductor muscles during running activity as well as the inter-subject variability. In addition, the study aimed to describe the typical EMG profile for the major lower limb muscles collected during running as well as the associated inter-subject variability.

2.8 Relationship between adductor activation and frontal plane kinematics during running

The first two studies (chapters 3-4) of this thesis are methodological in nature and seek to provide data to validate the proposed protocol for measuring adductor function during walking and running. In the following chapter (chapter 5), the adductor function during running is described. The final section is focused on understanding whether the proposed technique for measuring adductor activity can be used to address research question related to their function. Pelvic drop is an example of a kinematic patterns that has been associated with running injuries and which may be influenced by adductor muscle function. Following this idea, the relationship between adductor activation patterns (measured with EMG) and kinematic and kinetic characteristics during running will be explored. This aspect of thesis will test a new framework which could explain abnormal pelvic frontal movement.

Although running can lead to enhanced health benefits, musculoskeletal injuries are frequently reported by recreational runners, especially at the knee. Among knee injuries, patellofemoral pain (PFP) is a common problem with an incidence of 3–40% in active populations (Callaghan & Selfe, 2007; Messier, Davis, Curl, Lowery, & Pack, 1991; Neal,

Barton, Gallie, O'Halloran, & Morrissey, 2016). Blond and Hansen (1998) reported that 80% of PFP-affected people who had completed a rehabilitation programme still reported pain at their five-year follow up, and 74% had reduced their physical activity level as a result. This may be due to a failure to address the underlying factors that contribute to the development of PFP (Thomson, Krouwel, Kuisma, & Hebron, 2016), some of which may be related to coordination patterns of the hip and knee musculature.

Although the precise mechanism of anterior knee pain is still subject of debate, it is well accepted that increased patellofemoral joint stress must play a role. This increased stress will result from either patellar mal-tracking and/or patellar compression against the trochlea of the femur (Dierks, Manal, Hamill, & Davis, 2008). A number of studies have suggested specific mechanisms, local to the knee joint, which could lead to these stresses, such as increased Q-angle and a quadriceps muscle strength defect (Earl & Vetter, 2007; Fok, Schache, Crossley, Lin, & Pandy, 2013; Lenhart et al., 2015; Livingston, 1998; Pal et al., 2012). However, it has also been recognised that factors proximal to the knee joint can influence knee mechanics and result in PFP (Dierks et al., 2008). The most important of these factors are contralateral pelvic drop (CPD) and excessive hip adduction (Noehren, Sanchez, Cunningham, & McKeon, 2012; Willson & Davis, 2008a, 2008b; Willy, Scholz, & Davis, 2012), which may all, to some degree, be related to the coordination patterns of the hip adductor and abductor muscles.

In the frontal plane, at initial contact, the pelvis—in a healthy population—tilts laterally away from the stance limb (i.e. moves downward on the contralateral side). Following initial contact, there is a slight increase in this drop, after which there is a rapid elevation of the contralateral side of the pelvis, which results in the pelvis being elevated relative to the stance limb at toe-off (Figure 2-16). During early float, there is minimal frontal plane movement of the pelvis, and the cycle is then repeated on the contralateral leg (Preece, Mason, & Bramah,

2016). These normal mechanics are often altered in people with running pathology. For example, runners who have PFP tend to display excessive peak contralateral pelvic drop (CPD) (more than 3.5° above the mean value) and peak hip adduction (more than 3.5° above the mean value), relative to the control group (Willson & Davis, 2008a, 2008b). A combination of these motions will result in dynamic knee valgus and lateral patellar tracking. Such lateral patellar movement increases the loading forces on the lateral aspect of the patellofemoral joint (Besier, Gold, Delp, Fredericson, & Beaupré, 2008; Powers, 2003, 2010). In addition, Willy, Manal, Witvrouw, and Davis (2012) identified the presence of excessive CPD (up to 2.6°) in PFP-affected males as a cause of higher knee adduction. This will result in medial patellar tracking and ultimately increases medial patellofemoral joint stress.
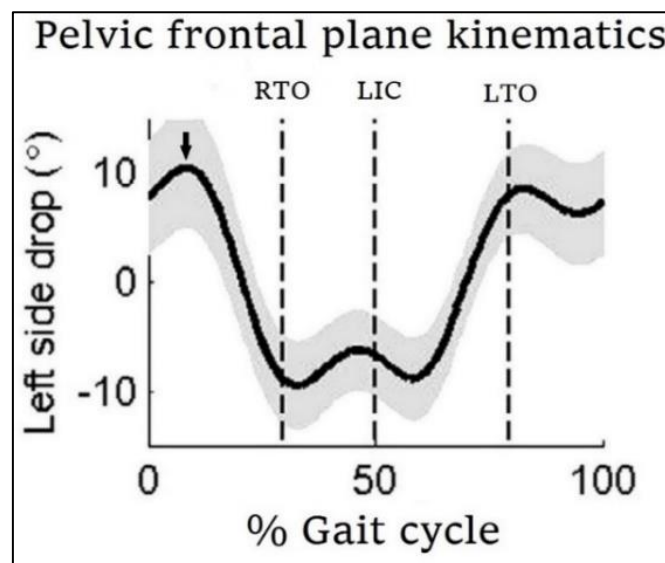
Figure 2-16: Ensemble average curve (across all [n=28] subjects), with standard deviation envelope, for the pelvis relative to the laboratory coordinate system frontal plane. Data is plotted from right initial contact (RIC) to the following RIC with the three vertical lines showing the timing of right toe-off (RTO), left initial contact (LIC) and left toe-off (LTO), respectively. The down arrow refers to the peak of left side pelvic drop (adopted from (Preece et al., 2016)).

CPD and excessive hip adduction have also been proposed as risk factors for other running related injuries (RRI) (Fredericson et al., 2000; Noehren, Davis, & Hamill, 2007; Pohl, Mullineaux, Milner, Hamill, & Davis, 2008). Recreational runners with tibial stress fracture,

or iliotibial band syndrome, tend to display greater hip adduction relative to healthy controls (Fredericson et al., 2000; Noehren et al., 2007; Pohl et al., 2008). Significantly, CPD has also been shown to be a risk factor for lower limb pathologies that are non-running related, such as osteoarthritis (OA). Park et al. (2010) and Hinman et al. (2010) observed the presence of excessive hip adduction angles in knee OA patients. It is possible that CPD, as well as being associated with other lower limb kinematic changes, may also influence kinetics. For instance, the magnitude of the hip adduction moment was increased at single leg support in those with knee OA (Astephen, Deluzio, Caldwell, Dunbar, & Hubley-Kozey, 2008). Likewise, Allison et al. (2016) found that individuals with gluteal tendinopathy exhibited greater hip adduction moment, which was associated with pelvic drop and contralateral trunk lean during walking. This indicates that CPD is an important kinematic pattern that needs to be studied in greater depth.

It has been suggested that pelvic drop is the result of weak hip abductor muscles (Fredericson et al., 2000). Theoretically, running is mostly a sagittal plane activity; therefore, muscles associated with the frontal plane could become weakened without cross-training or strengthening (Burnet, Arena, & Pidcoe, 2008). In support of this idea, Fredericson et al. (2000) measured hip kinematics following a six-week hip-strength programme with a focus on GMed strengthening. They found that increased strength of hip abductors was associated with a reduction in hip adduction angle in distance runners with iliotibial band syndrome. Interestingly, similar results were obtained for non-running related injuries, such as subjects with knee OA (Park et al., 2010). Previous studies have suggested that strengthening the GMed fosters increased control of thigh adduction tendencies, thereby minimising the valgus vector at the knee. However, other research does not corroborate this idea; for example, Willy and Davis (2011) and Baggaley, Noehren, Clasey, Shapiro, and Pohl (2015) found no correlation between hip abduction strength and either peak hip adduction or pelvic drop

during running. Moreover, they determined that there was no relationship between GMed isometric torque and frontal plane pelvic drop in recreational runners.

There are a number of reasons for the lack of agreement (described above) between studies into pelvic drop and muscle strength. Firstly, it is likely that the hip abductor muscles do not need to contract maximally to create sufficient force to counteract pelvic drop during running at a self-selected speed. Additionally, maximum isometric strength may not be the most relevant method to predict the performance of the hip during a dynamic weight bearing task such as running. Moreover, it is not clear whether muscles measured weak in static tests will necessarily exert insufficient force during dynamic activities, such as running. Furthermore, the other muscles crossing the hip, such as tensor fasciae latae, may have a role in compensating any reduction in the force generation of the hip abductor muscles (Baggaley et al., 2015). This lack of a theoretical link between isolated measures of abductor muscle strength and pelvic movements during running may explain the inconsistency of previous research. These inconsistencies also indicate that further research is required to understand the functioning of other hip muscles in order to fully understand individual variations in pelvic drop.

Adductor and abductor muscles both generate frontal plane hip movements. Moreover, the way the pelvis moves in the frontal plane is determined by a range of biomechanical factors, including the relative positioning of the foot, centre of mass and hip joint centre, as well as the activities of a range of muscle groups including the adductors and abductors. However, to date, previous research has focused on the abductor group, with minimal focus on the adductor group. Importantly, an over-activation of the adductor muscles could increase the adduction movement at the hip and therefore increase CPD. However, this has not yet been investigated. Therefore, further research is required to understand the potential links between

adductor activity and pelvic motion in the frontal plane. It is possible that the findings from such a study will provide new insight into the pathomechanics of patellofemoral pain, which in turn may play a critical role in injury prevention and management.

As discussed earlier, most researchers and therapists focused their attention on abductor muscles activity and strength as the main causes for pelvic drop in various types of pathologies. Interestingly, there has been no study investigated the activity of the adductor muscles during the stance phase of running, and how this activity affects frontal plane pelvic movement. Consequently, the current study aims to investigate the association between frontal plane movement and the EMG activity of the hip adductor muscles during the stance phase of running.

2.9 Summary and overview of the studies conducted throughout the thesis

The critical systematic review of the literature, presented early in this chapter, highlighted the potential importance of the adductor muscles but also the lack of studies investigating the activation patterns of these muscles during gait. A possible reason for this paucity of research is the difficulty in measuring these muscles. Therefore, the first aim of this thesis was to create a standard method for measuring the electromyographic activity of this muscle group using surface EMG electrodes (see the summary of the study 1, section 2.9.1 below). Following the methodological study, further work is required to ascertain whether this method is reliable across different testing sessions. Therefore, the second study aimed to explore the degree of consistency between EMG measurements (from lower limb muscles) collected during gait on two different occasions (see the summary of the study 2, section 2.9.2 below). With a standardised and reliable method for measuring the EMG for the adductor muscles, it is possible to design experiments to improve our understanding of the

biomechanical function of the adductor muscles during running. Therefore, the third study aimed at describing the EMG profile for the adductor muscles during running (see the summary of the study 3, section 2.9.3 below). Finally, the link between the pelvic drop (one of the most common risk factors linked to running related injuries) and the pattern of activation of the adductor muscles was be explored (see the summary of the study 4, section 2.9.4 below).

2.9.1 Summary and overview of Study 1

This study aimed at understanding the potential for using surface EMG to measure activity in the adductor muscle group. In order to achieve this aim, three separate but complementary experiments were conducted. In the first experiment, the relative movement of the adductor muscles under the skin was quantified at different hip joint angles. In the second experiment, the positions of the adductor muscles were quantified during an incremental isometric contraction in order to identify the effect of muscle contraction on the relative position of the muscle. The third experiment was designed to explore the relationship between adductor torque and the magnitude of the adductor EMG signals during ramped isometric contraction.

Experiment 1

Muscle movement relative to the skin during dynamic conditions is one of the factors that can influence the validity of EMG measurements. Therefore, the first experiment of the first study aimed at quantifying the movement of the adductor muscles (AL, Gr, and AM) relative to the overlying skin at different hip flexion/extension angles measured in standing position. This was achieved using ultrasonography. These angles mimic the leg movement during the walking and running tasks. For this purpose, 10 healthy participants were volunteered to join the study and they were asked to assume different hip angles position while their adductor muscles were imaged (for more information see chapter 3 experiment 1).

Experiment 2

Previous research found that the degree of muscle contraction can affect the location of the muscle relative to the overlying skin, consequently affecting the validity of the EMG measurements (Delaney et al., 2010; Rainoldi et al., 2000) . Therefore, the aim of the second experiment of the first study was to quantify movement of the adductor muscles in relation to the overlying skin using ultrasonography during incremental isometric contraction. The same group of subjects were asked to perform five levels of muscle contraction and ultrasound imaging was conducted to image the adductor muscle group during each level of contraction (for more information see Chapter 3 experiment 2).

Experiment 3

The relationship between the magnitude of EMG activity of the adductor muscles and their generated torque has not yet been reported in the literature. Therefore, this study focused on exploring the relationship between adductor torque and the magnitude of the adductor EMG signals during ramped isometric contraction. The identified locations for the adductor muscles in experiment 1 were used in this study for surface EMG electrode placement. Ten healthy participants participated in this study (for more information see Chapter 3 experiment 3).

2.9.2 Summary and overview of Study 2

The primary aim of this study was to understand EMG repeatability for the adductors while the secondary aim was to understand the repeatability of the other muscles during walking and running. This secondary aim was included to facilitate comparison of the results from this thesis with findings from previous studies. In this study, the activity of 12 lower limb muscles (the GMax, GMed, VMO, VLO, AL, Gr, AM, MHam, LHam, TA, MGastro, and LGastro) was measured in ten healthy recreational runners on two different occasions

separated by a minimum of one week. In addition, a number of normalisation techniques were carried, in order to understand the effect that normalisation could have on between-sessions measurement variability.

2.9.3 Summary and overview of Study 3

This study focused on describing the EMG profile for the adductor muscles during running task. The aim of the study was to describe the profile of this group of muscles during the human ambulatory task especially during running. The peaks of activation for each of these muscles were also identified. Additionally, the variability between individuals was describe in this study as well. This information was included to provide a comprehensive view of adductor function during running gait. The study will be discussed in Chapter 5.

2.9.4 Summary and overview of Study 4

This study aimed to investigate the association between the frontal plane movement and the EMG activity of the hip adductor muscles at specific point of the stance phase of running. A cross-section research design was used in this study and 25 recreational runners were recruited to participate. They performed a number of overground running trials during which the EMG activity for 12 lower limb muscles (six on each side) and kinematic data were collected. After full analysis of the data, the participants were assigned to one of two groups. The first included participants with low frontal plane pelvic movement while the second group included participants with high frontal plane pelvic movement. This study will be discussed in Chapter 6.

Chapter 3: Using ultrasound to monitor the relative position of the adductor muscles in order to validate EMG measurement

The extensive systematic review of the literature, presented in the previous chapter, highlighted the potential importance of the adductor muscles but also the lack of studies investigating the activation patterns of these muscles during gait. A possible reason for this lack of research is the difficulty in measuring these muscles. Therefore, the aim of this chapter is to establish a standard method from measuring the EMG activity of this muscle group using surface EMG electrodes.

3.1 Background

The hip adductor muscles play a critical role in both movement and stabilisation of the leg and pelvis during walking and running. It has been proposed that, rather than functioning to create an adductor moment during walking, the adductors may contribute to sagittal plane moment generation. For example, Perry (2010) proposed that, in addition to the recognised hip extensors, AM could have a role in hip extension in early stance phase. Another idea proposed by Grimshaw, Lees, Fowler, and Burden (2007) and Mann and Hagy (1980) is that hip adductors are continuously active throughout the running gait, and function to stabilise the pelvis with respect to the thigh. However, although a number of authors have proposed specific functions of the adductors during both running and walking, these assertions are often not supported by hard scientific data. These data would need to take the form of EMG data collected from electrodes attached on the skin, overlying the different adductor muscles (or using fine wire techniques), and interpreted in the context of the task/movement being performed.

A number of measurement problems currently limit measurement of the adductor muscles during dynamic tasks. Firstly, there is no clear guideline for surface EMG electrode placement for the adductor muscles. Moreover, the individual hip adductor muscles are close to each other at the upper medial aspect of the thigh (Watanabe et al., 2009). Thus, small errors in electrode placement could increase the possibility of cross talk, defined as picking up signals from an adjacent muscle rather than the muscle over which the electrode is placed. In addition, there is no consensus in the literature about the technique used to place EMG electrodes over the adductor. EMG electrodes can be placed over the Gr muscle using a palpation method because of its superficial position (Tsuji et al., 2015; Tsuji et al., 2012; Wall-Scheffler et al., 2010). However, the AM and the AL muscles lie within close proximity to the Gr, so electrode placement over these muscles using traditional approaches such as visual or palpation methods can be difficult (Watanabe et al., 2009). Another option is to use fine wire electrodes, which are implanted directly into the muscle (Mann et al., 1986; Montgomery et al., 1994). However, this invasive technique may not be appropriate for the adductor muscles, especially during dynamic movements such as gait when the medial borders of the inner thighs can sometimes make contact.

It is crucial to recognise that the electrode position is not stationary with respect to the underlying muscle, because there is relative movement of the muscle and overlying skin/subcutaneous fat when the lower limb is moved. Such relative movement between the muscle and skin can influence the validity of EMG measurement. Therefore, the first part of this study is focused on understanding whether movement of the skin relative to the underlying muscle could invalidate EMG measurement and will address two issues: recognition of the adductor muscles and quantification of muscle movement relative to the skin. Such movement could arise from two separate mechanisms. Firstly, when the muscles contract there could be movement of the muscle relative to the skin. The actions of contractile

elements cause the muscle to be displaced in a transverse direction (across the muscle) relative to the skin. This displacement appears to be caused by an increase in the transverse width of the muscle as it shortens and the lateral movement of activated fibres (Enoka, 2015). The second mechanism is that when the hip moves through a range of flexion-extension, there could be some movement of the muscle relative to the skin. Gait studies illustrated associated ranges of 15°-35° in walking and from 5°-50° in running (Novacheck, 1998). Such a large range of motion could lead to relatively large changes in the relative position of the electrodes. The associated movement of the skin and, subsequently, the electrode placement for the adductor muscles needs more investigation for different hip positions.

There is a need to develop an experimental methodology that can inform the precise way surface EMG electrodes are placed over the adductor muscles and can quantify the relative movement of the adductors during dynamic contraction. Such investigation could be done by using ultrasound (US) to identify the adductor muscle's location. With this approach, US could be used to monitor muscle border locations in various hip positions, which match directly different phases of gait cycle. Therefore, the first part of this study is focused on quantifying the movement of the adductor muscles, relative to the skin, at different hip joint angles in a standing position. As previously mentioned, when a muscle contracts, the muscle moves slightly relative to the skin to a new location, with the degree of displacement depending upon the amount of muscle contraction (Brown & McGill, 2010). Therefore, the second part of this study is focused on quantifying the movement of the adductor muscles, relative to the skin, at different levels of muscle contraction in a standing position.

The use of ultrasonography in EMG studies is currently a growing area of interest. To date, several studies have used US to define the anatomical boundaries and the superficial region of individual adductor muscles (Lovell et al., 2012; Watanabe et al., 2009). In addition to

being a useful tool for locating muscle boundaries, ultrasound has been used to measure muscle thickness (Thoirs & English, 2009), and the elongation of muscle and soft tissue structures during maximal (Alegre, Ferri-Morales, Rodriguez-Casares, & Aguado, 2014; Konrad & Tilp, 2014) and submaximal isometric contractions (Koppenhaver, Hebert, Parent, & Fritz, 2009). To date, only a few studies have used ultrasound to inform the placement of EMG electrodes over the adductor muscles (Lovell et al., 2012; Watanabe et al., 2009). Further, no studies have quantified the relative movement of the adductors during dynamic contraction. Therefore, more investigation is required to understand the influence of different levels of muscle contraction on the relative movement of the muscle in relation to the overlying skin.

As well as considering the relative movement between the skin and muscle, this study will explore the relationship between the force produced by the adductor muscles and their corresponding EMG signals. In order to interpret EMG measurements, there needs to be a clear relationship (possibly linear) between the magnitude of the EMG signal and the force produced. Therefore, this study will aim to quantify this relationship during a series of static (isometric) muscle contractions. Previous research has investigated the relationship between ramped isometric contractions and the torque produced by the muscles for a range of muscle groups (Beck et al., 2004; Coburn, Housh, Cramer, et al., 2004; Coburn, Housh, Weir, et al., 2004; Ebersole et al., 1999; Orizio, Perini, & Veicsteinas, 1989; Ryan, Beck, et al., 2008; Ryan, Cramer, Egan, Hartman, & Herda, 2008; Ryan et al., 2007). However, to date, there is no study exploring the relationship between the torque of a ramped protocol of isometric contraction produced by the adductor muscles and the magnitude of their corresponding EMG signals. Therefore, the EMG signal picked up during a ramp isometric contraction will be correlated with the force produced by the adductor muscles.

3.2 Aim of study

There were three separate but complementary aims to this study. In the first experiment, the relative movement of the adductor muscles under the skin was quantified at different hip joint angles. In the second experiment, the adductor muscle positions were quantified during a ramped isometric contraction in order to identify the effect of muscle contraction on the relative position of the muscle. In the third experiment, the relationship between adductor torque and the magnitude of the adductor EMG signals during ramped isometric contraction was explored.

3.3 Research questions

This study addressed the following research questions:

- Experiment 1: Is the position of the adductor muscles, relative to the skin, similar at various hip joint angles in a standing position?

- Experiment 2: Is the position of the adductor muscles, relative to the skin, similar while performing a ramped isometric contraction?

- Experiment 3: Is there a relationship between the torque produced by adductor muscles and the magnitude of the corresponding EMG signal during an isometric contraction?

3.4 Methodology
3.4.1 Research design

A repeated measures design was used to quantify the relative position of the adductor muscles in relation to the overlying skin, using ultrasound during a range of tasks. In addition, the same design was used to understand the relationship between the torque

produced by the adductor muscles and the magnitude of the corresponding EMG signal during an isometric contraction.

3.4.2 Participants and participant recruitment

Ten healthy male volunteers ranging from 18-40 years were volunteered to participate in the study. The study was limited to participants with a body mass index (BMI) below 25. The age group was selected to represent the young and middle-aged recreational population, as a low BMI is more likely to be found in this group. Before testing, the decision was made to exclude any data collected from participants with subcutaneous fat thickness more than 2 cm over the inner thigh as this could affect the EMG measurements. Fat thickness measurement was undertaken using the real time US system which displays the depth from the skin surface up to the superior surface of the desired muscle including the subcutaneous layer (Figure 3-1). The subcutaneous layer was measured in order to decide the eligibility of each subject to participate in the study. For all ten volunteers, fat thicknesses over the adductor muscles were less that the 2 cm threshold and therefore no participants were excluded. Across the subject, the mean fat thickness for AL was 1 ± 0.2 cm, for Gr was 0.7 ± 0.3 cm, and for AM was 0.7 ± 0.3 cm. With no prior data, it was difficult to afford a formal calculation of the required sample size. Therefore, the sample size was selected to be similar to that used in previous studies (Ishikawa, Komi, Grey, Lepola, & Bruggemann, 2005; Kay & Blazevich, 2009; Muraoka, Kawakami, Tachi, & Fukunaga, 2001) and should be sufficient to capture anatomical variation in muscle morphology across different individuals.
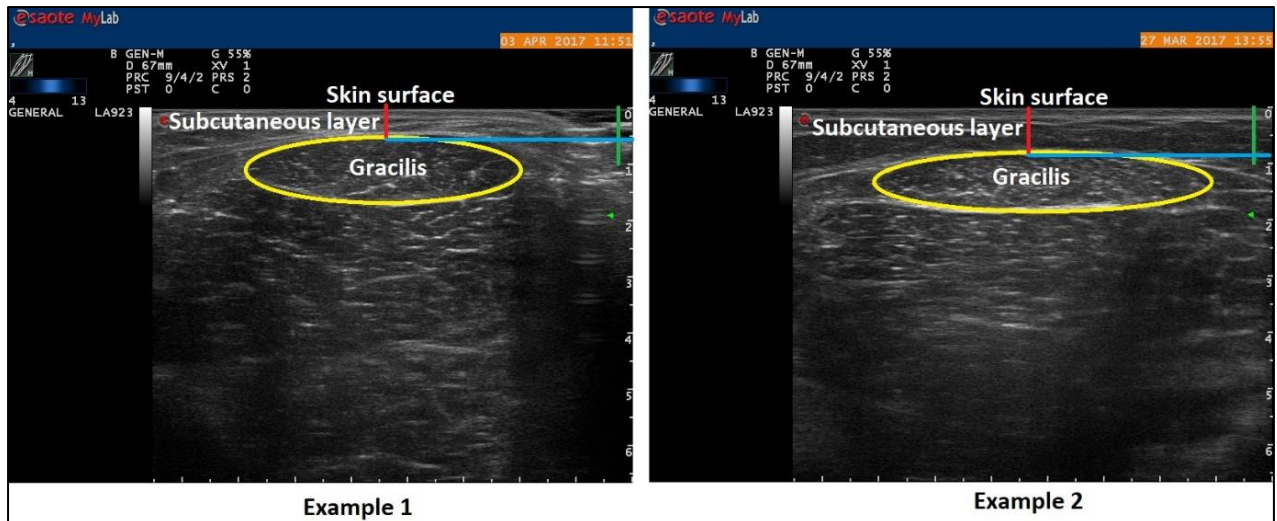
Figure 3- 1: The process of fat thickness measurement for two different participants of the gracilis muscle in which the vertical distance (red line) from the skin surface up to the superior border (the yellow oval shape) including the subcutaneous layer were measured using the scale on the top right of the figure (green line).

Other inclusion and exclusion criteria were used in addition to age and BMI. Subjects were needed to be free from any lower limb injury, especially in the groin and adductor area, for a minimum of six months before testing. Injury was defined as any musculoskeletal disorder that prevent the subject from performing his normal exercise habit. Also, subjects were needed to have no history of lower limb surgery. To ensure that subjects were recreationally active, they were required to have taken part in a minimum of 30 minutes of physical activity, three times a week, during the past six months, which included recreational and competitive sports. Participants were required to be able to perform maximum voluntary isometric contraction (MVIC) tasks comfortably. Volunteers who did not match the aforementioned criteria were excluded from the study.

The study was permitted by the University Research and Ethics Committee (see Appendix II) and all subjects gave written informed consent preceding their participation. Any subject who concurred the inclusion criteria and agreed to join the study was asked to avoid strenuous physical activity during the day of testing and the day before (Boyer & Nigg, 2004) to avoid

cumulative muscle fatigue that could lead to biased experimental results (Rainoldi et al., 2001).

3.4.3 Ultrasound procedure for imaging the adductor muscles

Ultrasonography, also recognised as diagnostic sonography or medical ultrasound, is an imaging procedure that can be used to view the structures of the body such as tendons, muscles, joints, vessels, and other internal body parts. This technique uses sound waves with a frequency higher than is audible to humans (>20,000 Hz). Ultrasound images are created by sending and receiving ultrasound into human tissues using a probe. Each body organ produces its own 'echo' which is, in turn, recorded and presented as an image for the user. Several kinds of images can be formed using sonographic equipment. The most widely-used type in the musculoskeletal field is a brightness mode (B-mode), in which a linear array of transducers simultaneously scans a plane through the body that can be viewed as a two-dimensional image on screen. This demonstrates the auditory impedance of a two-dimensional (2D) cross-section of the tissue. Usually, superficial structures such as muscles and tendons are imaged at a relatively high frequency (7-18 MHz). This frequency affords better axial and lateral resolution.

The ultrasound imaging technique has been extensively used as a valid and reliable tool in the field of human biomechanics for many purposes. It has been used to investigate the degree of change in the inner muscle structure during different tasks. For instance, it has been used to explore the change in fascicle lengths for gastrocnemius and soleus muscles during walking (Ishikawa et al., 2005), and to follow up the effect of a ballistic stretching training programme (Konrad & Tilp, 2014). Similarly, it has been used to estimate muscle mass at different body locations such as the upper arm, forearm, anterior and posterior thigh, and lower leg (Thoirs & English, 2009). Moreover, ultrasound has been used to guide the

insertion of fine wire EMG electrodes (Chapman et al., 2010; Murley, Menz, & Landorf, 2009). For the same purpose, it has also been used to guide surface EMG electrode placement during static (Lovell et al., 2012) and dynamic tasks (Watanabe et al., 2009). Despite the wide range of ultrasound applications in biomechanics, it has not been used to understand how the muscle moves relative to the overlying skin at different limb positions and also during muscle contraction.

Ultrasound machine set-up

A MyLab70 (Esaote, USA) ultrasound imaging system was used to perform the ultrasound measurements (described below) required to address most of the research questions in this study. After checking the recruitment criteria and signing the consent form, series settings on the ultrasound machine were adjusted in order to get a high-quality image and to ensure the consistency between measurements. Initially, personal details such as name and date of birth were entered on the start-up screen. Each participant was then given a specific identification number to facilitate the recalling process and to ensure participant privacy. On the same screen page, the musculoskeletal application and probe size were chosen. The linear array probe LA923 was used in this study as it was the most appropriate size to cover most of the inner thigh muscles, being 10 cm long (Figure 3-2).

Figure 3-2: The linear array probe LA923.

The control panel consists of four main components: the software keys section with dedicated LCD screens, the alphanumeric section (keyboard and ON/OFF button), the trackball, and a controls section (Figure 3-3). The function of the most important keys used in the current study is provided in Table 3-1.
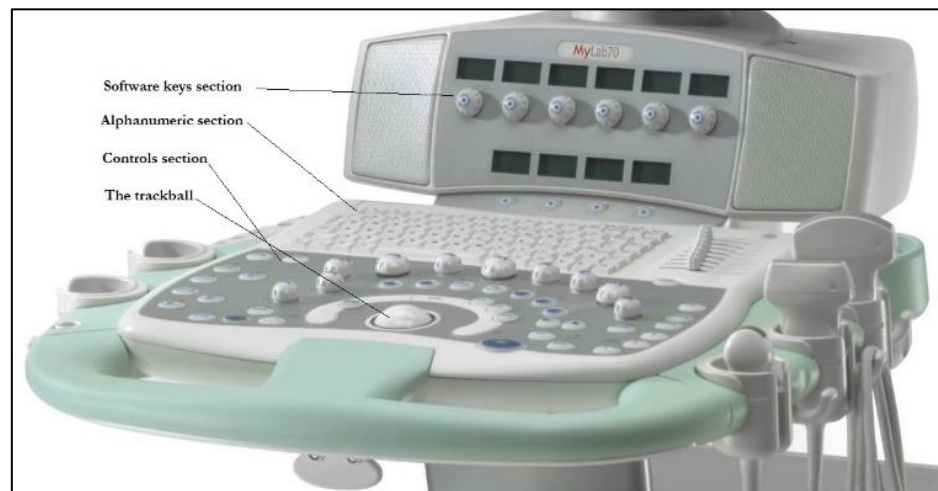
Figure 3-3: The control panel.

Fine-tuning of the ultrasound setting is an essential process for gaining high-quality images. Therefore, certain steps were taken in order to optimise image quality. Firstly, frequency was adjusted via two different settings. The first step was to set the frequency band (range), and a range of 7-18 MHz is regarded as suitable for imaging muscles and tendons. The second adjustment allowed fine-tuning of the frequency through a number of different settings: resolution-high (RES-H), resolution-low (RES-L), general mid-range (GEN-M), and penetration-medium range (PEN-M). Each of these settings is optimised for imaging a specific tissue at a specific depth in the body. Following guidance from a radiographer and through experimentation, it was found that using the setting GEN-M optimised image resolution of the adductor muscles (Figure 3- 4d). However, using other kinds of resolution e.g. resolution-high (RES-H), resolution-low (RES-L) results in poor quality image for the adductor muscles (Figure 3- 4a). Optimal frequency settings were used for all participants.

69

For each participant, image quality was optimised by using the setting described above and then by fine-tuning some additional settings. Firstly, the contrast was adjusted in order to maximise the visibility of the muscle borders (Figure 3- 4e). This contrast enhancement was performed to maximise the visibility of muscle boundaries (Figure 3- 4b). Secondly, the scanning zoom was set to focus the probe on a small area or to zoom out to look at a wider area. For the present set of experiments, this was done on individual basis depending on muscle geometry (i.e. the anterior and posterior borders of the Gr muscle needed be clearly seen on the screen). This resulted in an optimal definition of the boundaries and more regular shape of the muscle (Figure 3- 4f), however, (Figure 3- 4c) shows poor identification of muscle boundaries.

For each measurement, the probe was aligned in horizontal orientation at right angles to the skin and held with minimal pressure over the tested area. Also, the green light of the probe was oriented in order to face the examiner so that the left-hand side of the ultrasound image (see red symbol ꙫ on the left-hand corner of (Figure 3- 4 below) corresponded to the aspect of the muscle nearest the examiner. The US was prepared for the imaging process through these set-up procedures.
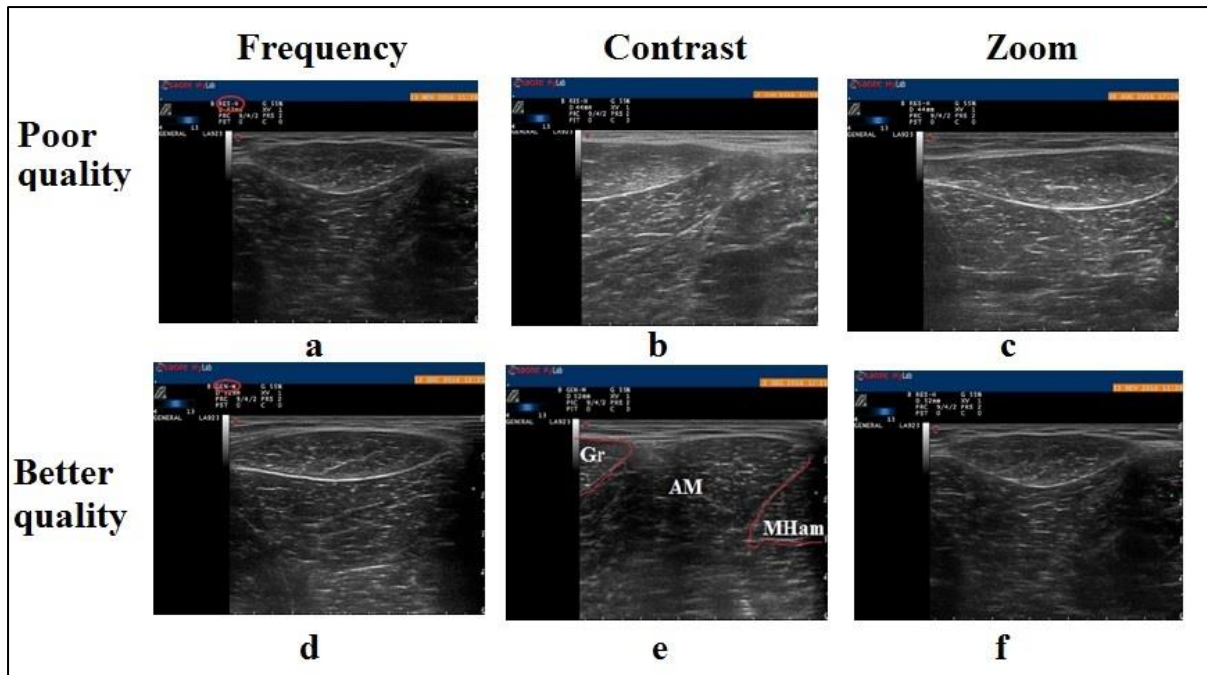
Figure 3- 4: Examples of poor and high-quality images. Images a-c demonstrate poor-quality images, while d-f represent high-quality images, which were used in the study.

Table 3-1: Summary of the most important controls section keys and their functions

Button	Function
Start End	This key opens and closes every exam.
PROBE	This key is used to select a different probe during the investigation.
Patient ID	This key is used to input or modify the patient's data during the exam.
B-MODE	This key re-activates a B-Mode image in real time when the system is used in any other mode.
✳	This key stops the current analysis or scan and puts the system in Freeze mode. To re-activate in real time, the same button is pressed again.
DEPTH ZOOM	The DEPTH-ZOOM multifunction key allows the examiner to change the scanning depth or to enlarge the 2D image.
Orientation	This key change right/left or left/right orientation, indicated by a symbol *e* that represents the probe led.
FREQ TEI	This knob allows the imaging frequency to be quickly changed (higher frequency to optimise resolution, or lower frequency to increase penetration).
IMAGE CLIP	During the exam, the operator can save both individual images and sequences (for systems having the clip licence) by using the IMAGE key, for the frames, and the CLIP key, for the sequences. The stored images and sequences are displayed as thumbnails at the bottom of the screen.
EXAM REV ARCHIVE REV	The EXAM REV key is used for accessing, at any time, the data stored during the current exam. To access the data archive, the examiner presses the ARCHIVE REV key.

Adductor muscles location

A pilot study was undertaken to accurately locate the adductor muscles in their most superficial position and to standardise the probe-positioning between participants. Initially, the distance from the greater trochanter of the femur to the lateral knee joint line was measured and 60-80% of this distance used as a guide to identify muscle position. The femoral lengths of the tested participants were ranged from 38-42 cm. The location of the adductor muscles was then investigated at three different levels: 60, 70, and 80% of femur length, respectively, using the ultrasound probe (Figure 3-5). This process involves putting a water-soluble transmission gel over the participant's skin and then using a probe to image the underlying muscle structures. At the level of 60%, it was easy to locate the Gr and AM but the AL was poorly defined (see Figure 3-5b, f, and j). At the level of 70%, the width of the Gr and AM slightly narrowed compared to the former level (see Figure 3-5g and k). The AL appeared at this level but with narrow width (see Figure 3-5c). At the level of 80%, the edges of the Gr and AM were much closer than in the aforementioned levels (60% and 70%) and the AL was clearly defined at this level (see Figure 3-5d, h and l).

This pilot study involved five different volunteers and the results were consistent for all measurements. Therefore, given this preliminary work, the protocol summarised in Table 3-2 was adopted for all subsequent ultrasound measurements of the adductor muscles.

Table 3-2: Summary of the findings of the pilot study

Muscle	The best superficial location of the muscle
Adductor longus	At 80% of the femur length
Gracilis	At 60% of femur length
Adductor magnus	At 60% of femur length
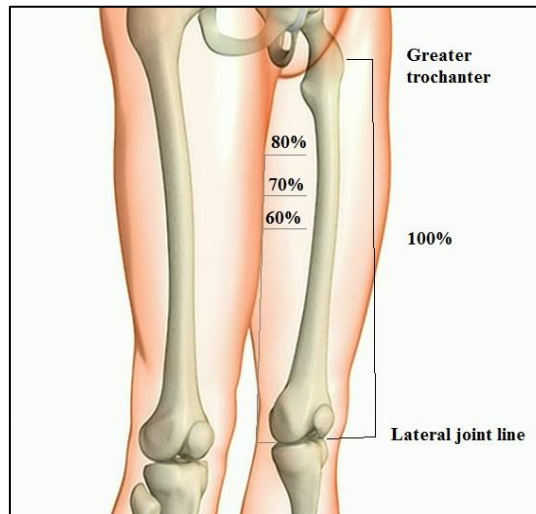
Figure 3-5: Different levels to approach the adductor muscles.
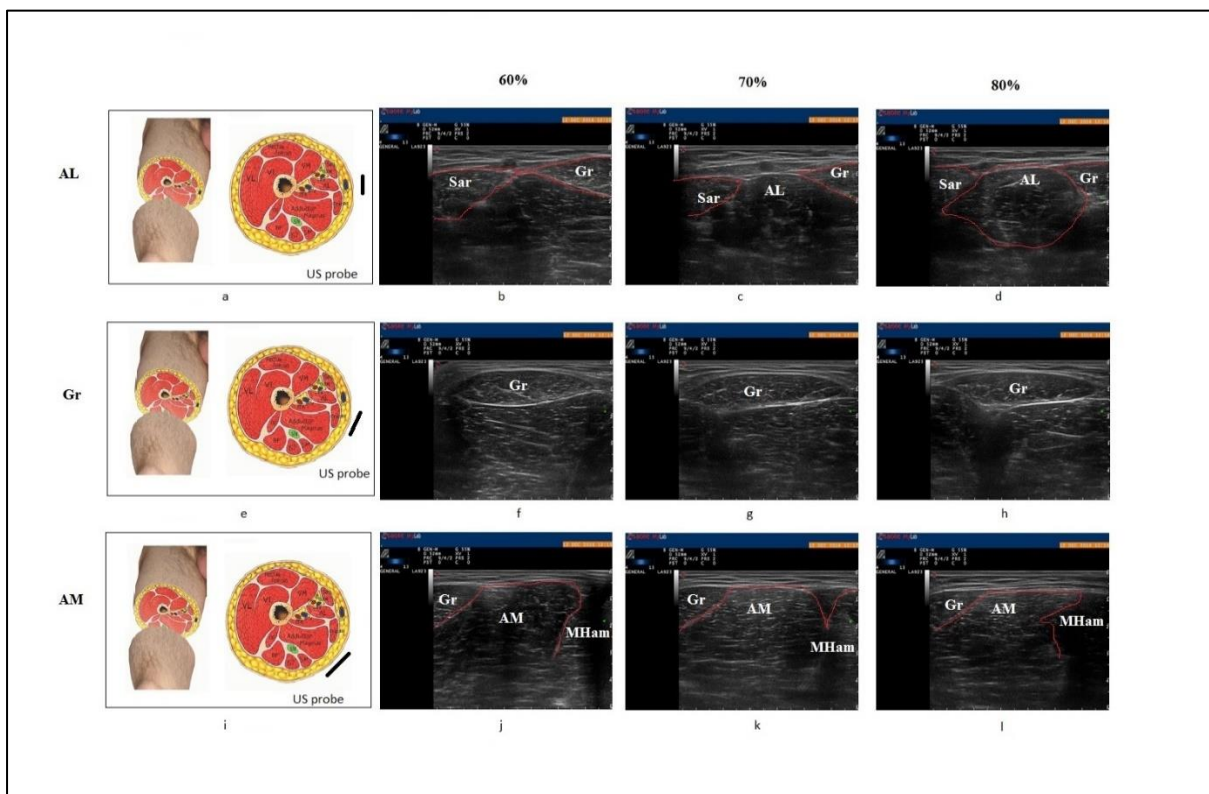
Figure 3-6: This figure displays a cross-section of the adductor muscles at different levels of the ultrasound scan, in order to locate the most superficial portion of the adductor muscles at these levels. The adductor muscles were scanned at 60, 70, and 80% of the femur length. AL: adductor longus; AM: adductor longus; Gr: gracilis; Sar: Sartorius; MHam: medial hamstring.

74

3.4.4 Experimental data collection for Experiment 1

Aim: The main objective of this experiment was to quantify the movement of the adductor muscles, relative to the overlying skin, at different flexion/extension positions, which correspond to the range associated with walking and running.

Research design: A repeated measures design was used to quantify the relative position of the adductor muscles in relation to the overlying skin using ultrasound while assuming different hip flexion/extension movements. Ten healthy runners were recruited to participate in this experiment.

After signing the consent form, each participant was requested to change into their shorts and a comfortable t-shirt. First, anthropometric data such as body mass and height were measured using universal weight and height scales. As described in the pilot study (see Table 3-2), the position of the adductor muscles, on the medial aspect of the thigh, was set at 60% and 80% of the total femur length. The Gr and AM muscles were located at 60% while the AL muscle was located at 80%. Next, the edges of the Gr, septum between MHam and AM, and medial border of the sartorius muscle were marked using a water-based (non-toxic) marker pen (Figure 3-7a and b) and the skin marked appropriately. Then, the centre of each muscle was marked (at the 60/80% longitudinal position) and these points were used as reference points for all subsequent experiments. Similarly, the centre of the ultrasound probe was marked as well.

In order to achieve the aim of the first experiment, the centre of the ultrasound probe was aligned over each reference point to monitor the relative movement of each muscle at a number of hip angles (0°, 20°, 40° of hip flexion, and 20° hip extension) (Figure 3-8). The reason for choosing these angles is that these hip angle degrees mimic the normal range of motion during walking and running tasks (Figure 3-9) (Novacheck, 1998; Preece et al.,

2016). When using the ultrasound, the participant was asked to stand at 0° hip extension and

an ultrasound image was taken for each of the adductor muscles at this angle. The participant

was then asked to assume the other hip angles and the same imaging process was used at each

angle. At each hip angle, one US image was taken for each of the adductor muscles. A

transparent plastic goniometer (Baseline® Plastic Goniometers) with a 360-degree head and

12-inch arms was used to confirm each hip joint angle (Figure 3-10). The participant was

encouraged to take one-minute rest between the ultrasound imaging measurements at

different hip angles.
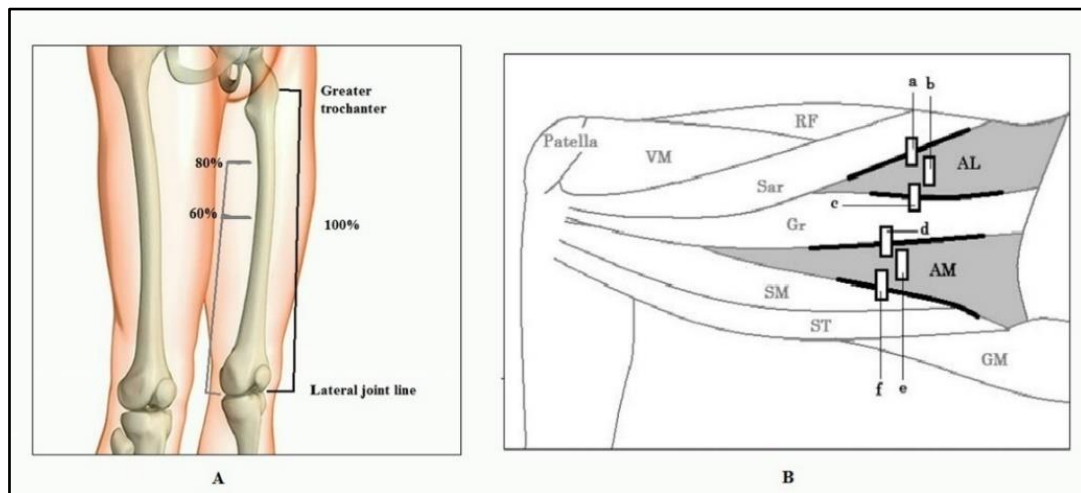
Figure 3-7: The process of locating the adductor muscles and marking their edges. AL: adductor longus; AM: adductor longus; Gr: gracilis; Sar: Sartorius; SM: semimembranosus; ST: semitendinosus. Letters a-f represents the probe position.
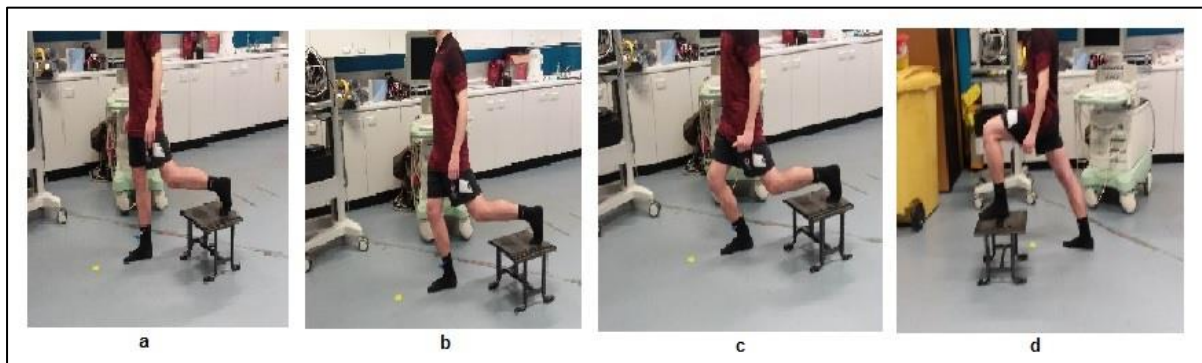
Figure 3-8: Testing positions for Experiment 1, during which the examiner conducted a series of ultrasound images for the tested group of muscles of the right lower limb; a. natural hip position, b. $20^\bullet$ hip flexion, c. $40^\bullet$ hip flexion, and d. $20^\bullet$ hip extension.
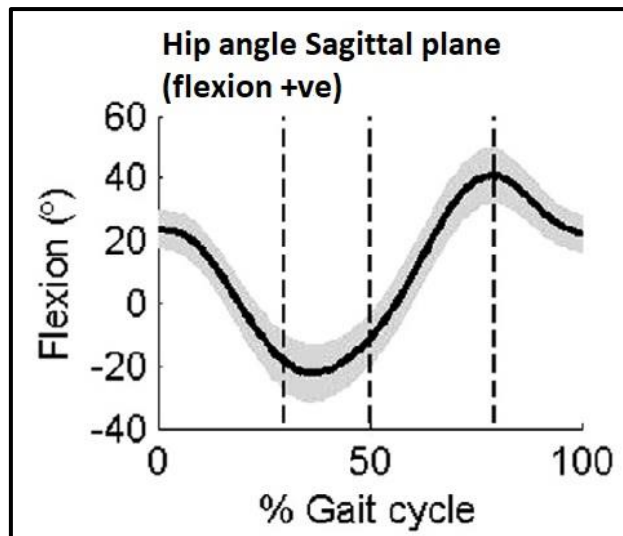
Figure 3-9: Sagittal plane movement of the hip during running (Preece et al., 2016).
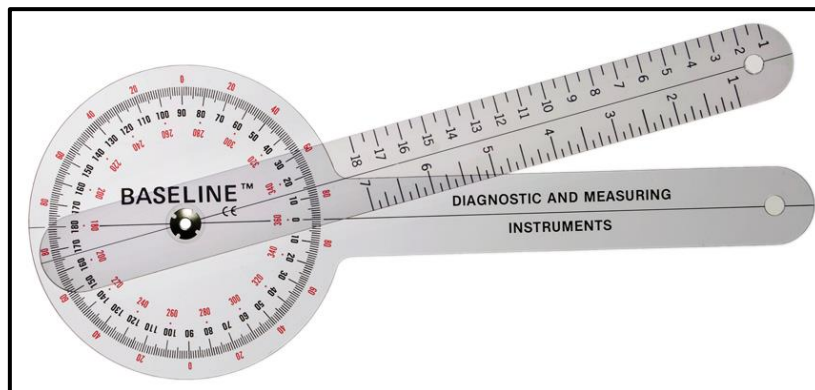
Figure 3-10: The transparent plastic goniometer.

3.4.5 Experimental data collection for Experiment 2

Aim: the main aim of this experiment was to investigate the relative movement of the adductor muscles in relation to the overlying skin while performing ramped isometric contraction.

Research design: A repeated measures design was used to quantify the relative position of the adductor muscles in relation to the overlying skin using ultrasound while performing ramped isometric contraction. The participants who participated in the first experiment were asked to join in this experiment as well.

In order to achieve the aim of the second experiment, the centre of the ultrasound probe was aligned over each reference point to monitor the relative movement of each muscle during different levels of contractions. The adductor muscles were imaged separately while performing the ramped protocol of contraction. A rest period of 2-3 minutes was given to the participant between the set of images for each adductor muscle. As previously mentioned in this chapter, in order to identify the degree of muscle displacement relative to the skin for each adductor muscle, the probe of ultrasound was placed over the same predefined mark on the skin, the reference point for each muscle (experiment 1), during all ultrasound imaging.

The torque produced during this isometric contraction was measured using the Biodex System 3 isokinetic dynamometer (Biodex Medical Systems, Shirley, NY). Isometric hip adduction strength was measured for each participant using a protocol described by Brent, Myer, Ford, Paterno, and Hewett (2013).The participant was instructed to stand facing the dynamometer head (Figure 3-11). The dynamometer head was aligned in parallel with the frontal plane of the body, with the axis of rotation of the dynamometer aligned with the centre of hip rotation. The tested limb was secured to the attachment arm with a custom strap and resistance pad extending from the attachment arm positioned immediately superior to the knee. The participant was instructed to grasp the top of the dynamometer head, aiming to minimise movement of the torso. The test began with measuring the maximum isometric contraction of the adductor muscles (averaging of three maximum trials), then, the participant was encouraged to produce 20, 40, 60, and 80% of that maximum contraction. Each participant could see his level of muscle contraction on a computer screen, which was considered biofeedback for his contraction level. Simultaneously, each of the adductor muscle was imaged separately at each level of muscle contraction using the ultrasound imaging system. At each level of muscle contraction, one US image was taken for each of the

adductor muscles. A rest period of 2-3 minutes was given to the participant between the set of images for each adductor muscle.

A number of approaches were used in this study to ensure that participants performed the maximal muscle contraction. These included a detailed explanation to the participant of the importance of giving the maximum effort during measurement. In addition, consistent verbal encouragement was used to motivate participants and visual feedback was used to show participants how hard they were pushing against the fixed resistance. This feedback essentially gave them something to aim for. Finally, a rest period of 30 seconds was given between each muscle contraction to allow recovery after each test.

It was not possible to directly synchronise the Biodex System with the Direct Transmission System used to collect the US data. Therefore, ultrasound images were collected separately at each force level. Specifically, the examiner aligned the US probe over the reference point, after which, the participant was instructed to produce the desired level of muscle contraction then hold this contraction level for a minimum of 2sec. During this period (2sec), the US image was taken for one of the tested muscles. This process was repeated throughout the ramped protocol of isometric muscle contraction for each of the adductor muscles in turn.

Figure 3-11: Testing position during isometric adduction contraction.

3.4.6 Experimental data collection for Experiment 3

Aim: The main objective of this experiment was to explore the relationship between the torque produced by the adductor muscles and their EMG magnitude during incremental isometric contraction.

Research design: A repeated measures design was used to explore the relationship between the torque produced by the adductor muscles and their EMG amplitude. The participants who participated in the first experiment were asked to participate in this experiment as well. Participants were asked to perform the third experiment before the second in order to avoid the effect of fatigue on the EMG signals.

Equipment

EMG data was collected using a Direct Transmission System with 16 channels (Noraxon USA Inc., model 586 Tele Myo DTS Desk Receiver) (Figure 3-12a). The DTS sensors (model 542) were used and EMG lead (542AP) set was inserted into each EMG probe

(Figure 3-12b), a disposable adhesive Ag/AgCl EMG electrode shaped in a figure-of-eight, and measuring 2.2x4 cm, with two 1 cm in diameter conductive circles and 2 cm separating each electrode. A computer program (Model 131 MyoResearch-XP) was the software used. EMG data were sampled at 3000 Hz. Muscle activity from each participant was recorded for three muscles: AL, Gr, and AM. The standard guidelines for reporting EMG data states that the surface EMG power spectrum ranges from 5 Hz and 500 Hz (Hermens et al., 1999; Stegeman & Hermens, 2007). Therefore, a sampling frequency of 3000Hz was deemed appropriate. The torque produced during the isometric contraction was displayed on a computer screen using the biofeedback mode in the Biodex System 3 isokinetic dynamometer (Biodex Medical Systems, Shirley, NY).
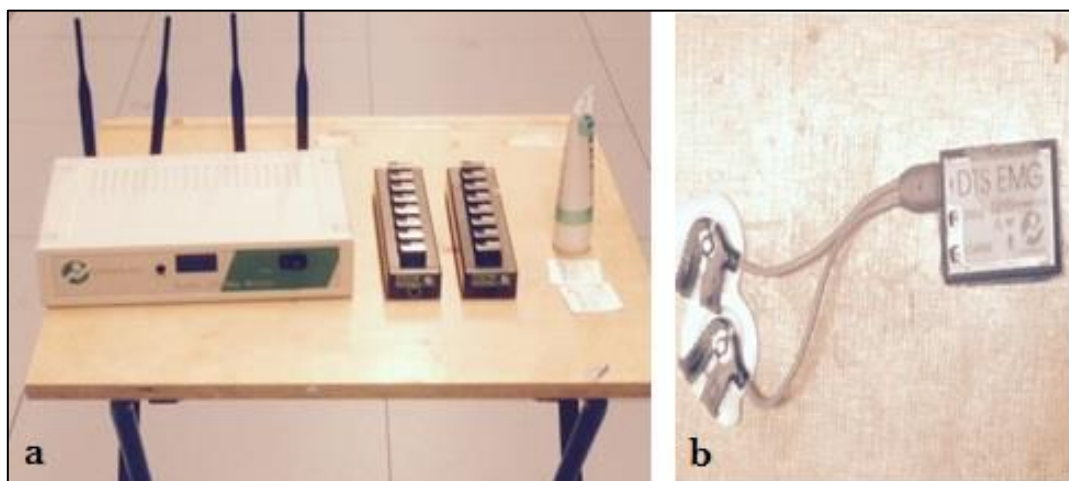
Figure 3-12: The EMG capture system: a. Direct Transmission System with 16 channels, b. the DTS sensors.

Skin preparation is a fundamental stage in collecting a high-quality EMG signal. This step was required to reduce artefact-causing interference, allow the electrode to stick firmly over the skin, and therefore to improve the signal quality. Initially, the skin over the predefined reference points (for more information see the experimental data collection for Experiment 1) was shaved using a disposable razor, then a special abrasive skin preparation (Nuprep Gel) was applied to the electrode site with a gauze pad to remove the dead skin. Thereafter, the skin area was cleaned with 70% isopropyl alcohol and left for two minutes to dry. Finally,

self-adhesive Ag/AgCl bipolar dual surface electrodes were placed over the prepared sites in line with the muscle fibres (Figure 3-13).

The adductor muscle electrode was placed over the skin on the inner side of the thigh, based on the findings of the pilot study (see Table 3-2). For all tested muscles, the recording electrodes were connected to transmitters and stuck over the skin next to the corresponding electrode. A final step before data collection is the signal checking process. For this process, the participant was instructed to completely relax and then to maximally contract a muscle. During both stages, the signal-to-noise ratio (SNR) was examined. A complete noise-free recording is impossible, therefore small amplitude spikes or random nature may be visible, but they should not exceed 10–15 microvolts. The ideal averaged baseline noise should be between 3–5 microvolts (Konrad, 2005). As walking and running EMG signals were typically in excess of 100-200 microvolts, a 10 microvolts noise ensured a signal to noise ratio of at least 10-20, which is well above that recommended for EMG data collection by the EMG system manufacturers retrieved from (https://www.delsys.com/products/software/emgworks/sqm/improve/). However, in most cases the signal was well above this level, ensuring a very high signal to noise ratio.

Once the signal checking process was finished, the electrodes and transmitters were secured with a crepe bandage (4 cm x 2.5 m) and athletic tape to minimise any movement artefact. Finally, the participant was asked to stand facing the dynamometer to be ready for data collection.
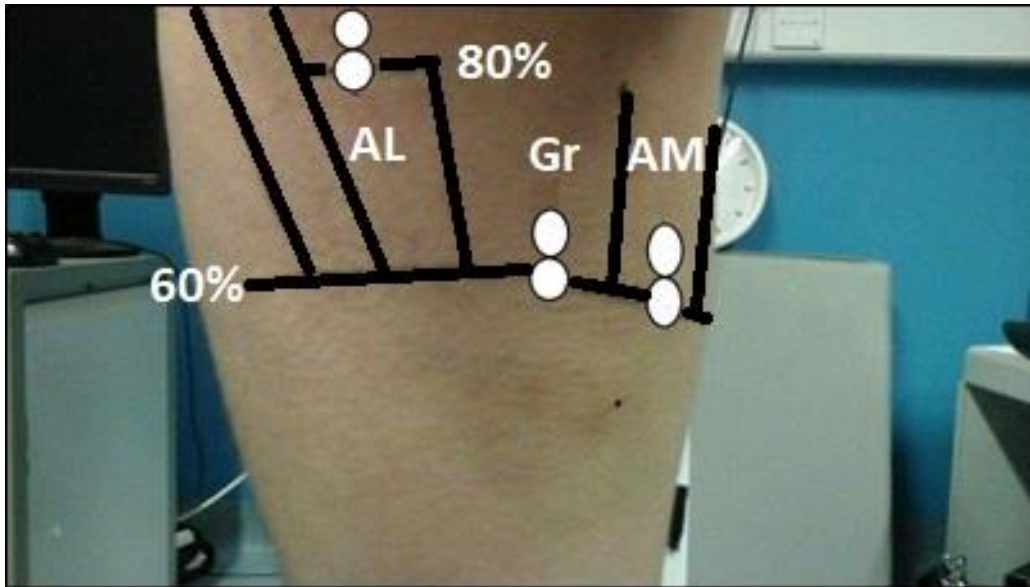
Figure 3-13: The location of surface EMG electrodes for the adductor muscles.

In order to explore the relationship between the force produced by the adductor muscles and the magnitude of the corresponding EMG signals, participants were asked to perform the same incremental protocol of isometric contraction performed in the second experiment during which EMG signals were recorded for this group of muscles. The torque produced during this isometric contraction was measured using the Biodex System 3 isokinetic dynamometer (Biodex Medical Systems, Shirley, NY. All muscle torque values were collected in Nm (Newton Meters) and later normalised to each participant's body mass (Nm/kg). Normalisation to body mass allowed more accurate comparison between participants and previous findings from the literature.

Similar to the second experiment, the test began with measuring the maximum isometric contraction of the adductor muscles (averaging of three maximum trials), then, the participant was encouraged to produce 20, 40, 60, and 80% of that maximum contraction. A computer monitor, as a biofeedback method, was positioned in the participant's field of view to assist him in achieving the required level of muscle contraction. Simultaneously, one second of the EMG signals for the adductor muscles were recorded during the incremental protocol. It was

not possible to directly synchronise the Biodex System with the Direct Transmission System used to collect the EMG data. Therefore, EMG data were collected separately at each force level. Specifically, the participant was instructed to reach the desired level of muscle contraction first and then held this contraction level for a minimum of 5 sec. During this period (5 sec), the examiner recorded the EMG amplitudes. This process was repeated throughout the ramped protocol of isometric muscle contractions.

Order of testing

There is a very small chance that the order of the isometric contraction tests may be influenced by order. Therefore, the order of the isometric contraction tests, described above, was randomised (Appendix III).

3.4.7 Data analysis for Experiment 1

The first experiment was aimed at quantifying the relative movement of the adductor muscles relative to overlying skin while assuming various hip joint angles in standing position. For each hip joint positions (0°, 20°, 40° flexion, 20° extension), the ultrasound probe was aligned over the reference point for each muscle, a point which represents the place of the surface EMG electrode for each muscle, to produce a series of images. The different images (corresponding to each hip joint position) were vertically aligned in an excel sheet (Microsoft Office Excel, 2016) (see Figure 3-15). Two vertical lines were then drawn over the central portions of the grouped images for each hip joint angle. These lines represented the surface EMG electrode boundaries over the skin. Thereafter, these images were saved and re-opened by Image J software.

Muscle dimensions were measured from scans off-line using Image J software (available at: http://rsb.info.nih.gov/ij/docs/index.html). The first step in this process was to convert the

pixel scale into millimetres. This was achieved by measuring a known distance (10 mm) on

the US image and then adjusting the scale appropriately within the image J (Figure 3- 14 step

1). Then, the muscle boundaries were marked from each side of the electrode borders (the

two vertical lines) (Figure 3- 14 step 2). Thereafter, the distance was measured from the

muscle border to its corresponding side of the vertical line (e.g. from the right border of the

muscle to the nearest vertical line) (Figure 3- 14 step 3). The image J software then generated

a new window with this distance in millimetres (Figure 3- 14 step 4). Each distance was

calculated and represented as right and left distance. This process was repeated for all images

of each muscle for each participant (Figure 3-15). This procedure was used to quantify the

minimum and maximum right and left distances for each muscle at the different hip angles.

All demographic data, and descriptive statistics including the minimum, maximum and

average right and left distances during different hip angles for all adductor muscles, are

provided (Table 3- 3).
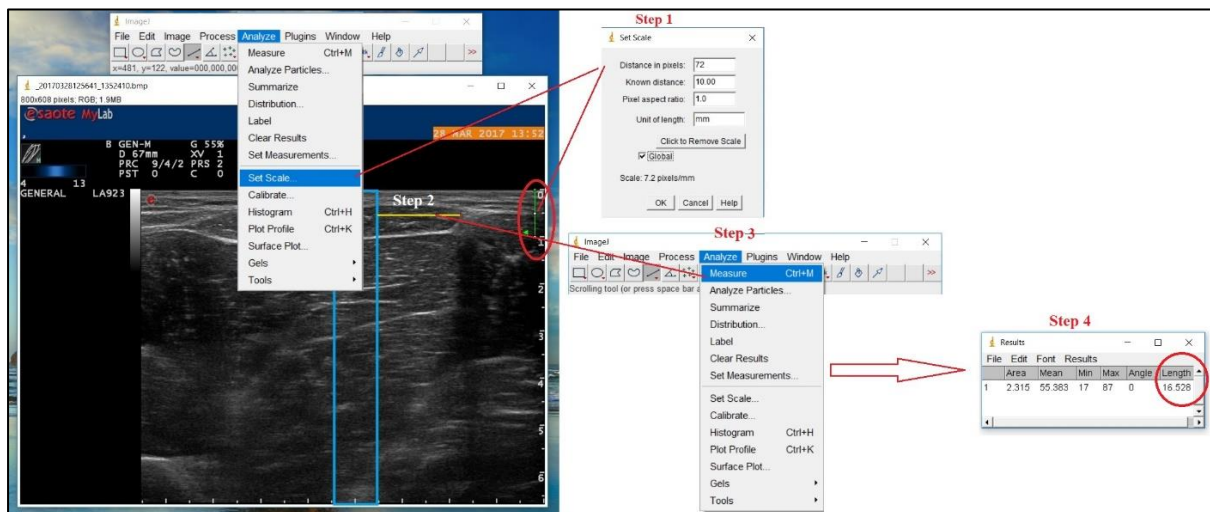
Figure 3- 14: The scaling process for the US image to convert the pixel measurements to a millimetre scale. The green vertical line, on the top right of the US image, is used to set and convert the scale from pixel to millimetre (step 1). The two vertical blue lines represents the surface electrode borders. Step 2 in which the distance from the muscle border to its corresponding side of the vertical line (electrode borders) was identified (the yellow line). Step 3 is used to measure the distance of step 2 with the new measurement unit (millimetre). Step 4 in which the measured length was achieved (the red circle).
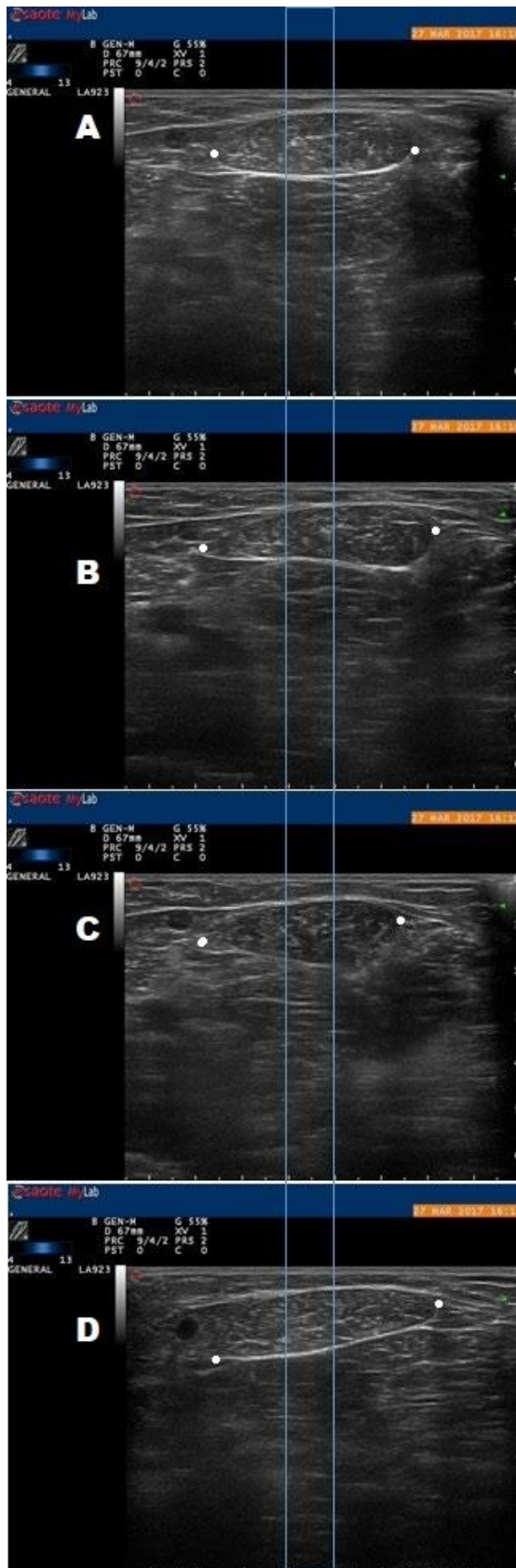
Figure 3-15: The ultrasound images for the gracilis muscles at various hip joint angles: A. 0°, B. 20°, C. 40° of hip flexion and D. 20° hip extension. The vertical lines represent the width of surface EMG electrodes.

3.4.8 Data analysis for Experiment 2

The objective of the second experiment was to quantify the relative movement of the adductor muscles in relation to the overlying skin while performing a ramped isometric contraction. Therefore, using the same approach as described for the first experiment, the distance between the edge of the muscle and the EMG electrode boundary (represented by two vertical lines) was quantified as right and left distances for each adductor muscle during each level of isometric contraction (20, 40, 60, 80, 100% MVIC) (Figure 3-16A-E). The same software, Image J, was used to measure the right and left distances. Similar to Experiment 1, descriptive statistics including the minimum, maximum, and average right and left distances during different levels of isomeric contraction for all adductor muscles are provided.

Figure 3-16: The ultrasound images for the gracilis muscles at different percentage of MVIC %: A. 20%, B. 40%, C. 60%, D. 80%, and E. 100%. The vertical lines represent the width of surface EMG electrodes.

3.4.9 Data analysis for Experiment 3

The aim of the third experiment was to investigate the relationship between the torque produced by the adductor muscles and their EMG signals during ramped isometric contraction. To analyse this data, a linear regression approach with standard errors adjusted for clustering was used. This statistical technique is able to deal with the repeated measures for each participant and was constructed to provide a p-value for the test of a linear relationship and to provide a percentage increase in EMG activity for each 1% increase in torque. Given that it was not possible to get every individual to contract their muscles at precisely the target levels of 20,40,60 and 80% of MVIC, the linear regression approach is appropriate as it can deal with the variation in ramped contraction values of MVC (as an independent variable) for each individual. The level of statistical significance was set at $P < 0.05$. To ensure the assumptions of regression were met the outcome was log-transformed in each instance. This ensured homogeneity of variance and that the residuals were normally distributed. Standard errors were adjusted for clustering at the subject level. The statistical analysis was carried out with Statistical Package for the Social Sciences (SPSS) (IBM SPSS Statistics 24), and graphs and tables were produced using Matlab (Mathworks, Matlab 7.10.00, R2010a) and Excel (Microsoft Office Excel, 2007).

3.5 Results
3.5.1 Results for Experiment 1

Ten healthy male subjects were volunteered to join the current study. The mean age of the athletes was 28.9 ± 7.78 years, mean height was 1.74 ± 0.05 m, mean weight was 70.2 ± 7.35 kg, and mean BMI was 23.15 ± 1.47 kg/m^2. The mean fat thickness for AL was 1 ± 0.2 cm, for Gr was 0.7 ± 0.3 cm, and for AM was 0.7 ± 0.3 cm. The descriptive statistics that include the minimum, maximum and average right and left distances measured from the electrode

border to the corresponding muscle border at different hip joint angles for all adductor

muscles are presented in Table 3- 4.

Table 3- 3:Right (Rt) and left (Lt) distances (±SD) of adductor muscles measured from the centre of marked skin to the corresponding muscle border on ultrasound images at different hip joint angles. All values are presented in millimetres.

Muscle	Hip joint angles	Min distance (mm)		Max distance (mm)		Mean distance (mm		Total muscle width (mm)
		Rt	Lt	Rt	Lt	Rt	Lt	
AL	0º	8	9	22	21	16 ± 5	15 ± 5	41 ± 11
	20º flexion	10	11	31	23	18 ± 7	16 ± 4	44 ± 10
	40º flexion	9	7	28	29	18 ± 8	18 ± 6	46 ± 13
	20º extension	6	10	23	22	16 ± 6	14 ± 4	40 ± 9
AM	0º	6	7	26	36	16 ± 7	19 ± 9	46 ± 14
	20º flexion	6	6	36	36	18 ± 11	18 ± 11	46 ± 16
	40º flexion	6	6	35	36	19 ± 10	18 ± 9	47 ± 15
	20º extension	7	7	34	32	19 ± 11	17 ± 9	45 ± 13
Gr	0º	6	7	24	25	18 ± 6	18 ± 7	46 ± 13
	20º flexion	6	6	26	36	16 ± 7	21 ± 9	47 ± 15
	40º flexion	6	6	34	30	17 ± 9	19 ± 8	45 ± 15
	20º extension	6	6	33	28	18 ± 9	19 ± 8	47 ± 14

SD: Standard Deviation; AL: Adductor longus; AM: Adductor magnus; Gr: Gracilis. Rt: The distance from the right side; and LT: The distance from the left side.

3.5.2 Results for Experiment 2

The descriptive statistics that include the minimum, maximum, and average right and left

distances measured from the measured from the electrode border to the corresponding muscle

border at different levels of isometric contraction for all adductor muscles are presented in

Table 3- 4.

Table 3- 4: Right (Rt) and left (Lt) distances (±SD) of adductor muscles measured from the centre of marked skin to the corresponding muscle border on ultrasound images at during incremental isometric contraction. All values are presented in millimetres.

Muscles	% of Muscle contraction	Min distance (mm)		Max distance (mm)		Mean distance (mm		Total muscle width (mm)
		Rt	Lt	Rt	Lt	(Rt)	(Lt)	
AL	20%	6	6	22	24	12 ± 5	15 ± 6	37 ± 9
	40%	7	5	25	20	14 ± 6	12 ± 4	36 ± 9
	60%	10	6	19	26	14 ± 4	13 ± 5	37 ± 8
	80%	9	5	17	21	12 ± 3	14 ± 5	36 ± 6
	100%	9	6	18	25	13 ± 3	14 ± 6	37 ± 7
AM	20%	6	7	27	28	18 ± 7	17 ± 7	45 ± 13
	40%	5	5	29	28	17 ± 9	15 ± 8	43 ± 13
	60%	6	5	28	29	16 ± 9	16 ± 9	43 ± 15
	80%	5	3	29	33	16 ± 8	17 ± 11	43 ± 15
	100%	6	3	29	30	17 ± 9	16 ± 8	43 ± 15
Gr	20%	8	9	27	28	18 ± 6	20 ± 7	47 ± 12
	40%	6	8	29	28	19 ± 7	18 ± 7	48 ± 13
	60%	9	6	31	25	19 ± 7	17 ± 6	46 ± 12
	80%	6	6	34	28	20 ± 9	18 ± 8	48 ± 14
	100%	8	4	27	25	15 ± 7	17 ± 7	42 ± 12

SD: Standard Deviation; AL: Adductor longus; AM: Adductor magnus; Gr: Gracilis; Rt: The distance from the right side; and LT: The distance from the left side.

3.5.3 Results for Experiment 3

A set of representative hip adductor isometric torque measurements (bottom) during 20%, 40%, 60%, and 80% of the MVC and their corresponding EMG signals (top) from the AL, AM, and Gr muscles are shown in Figure 3-17. Figure 3-18 shows individual plots of RMS EMG amplitude against isometric torque for the three separate adductor muscles in two separate individuals. Note that in these examples, the participants were able to achieve the target torque levels of 20, 40, 60 and 80% MVIC. The regression models demonstrated a linear fit between the torque and the muscle activity ($p<0.001$) for each of the three adductors and also the combined adductor activity. These models showed that isometric torque significantly affected muscle activity where, increasing isometric torque by 1% increased the AL, AM and Gr activity by the percentages shown below (Table 3- 5).

Table 3- 5: Fit of the linear regression model (where b is the slope of the line) and percentage increase in muscle activity for every 1% increase in torque for adductor longus (AL), adductor magnus (AM), gracilis (Gr) muscles and the combined EMG amplitudes of all three adductor muscles

Muscle	The regression coefficients (Estimate b)	P-value for fit of the linear regression model
AL	4.0	<0.001*
AM	3.3	<0.001*
Gr	3.8	<0.001*
Combined Adductors	3.4	<0.001*
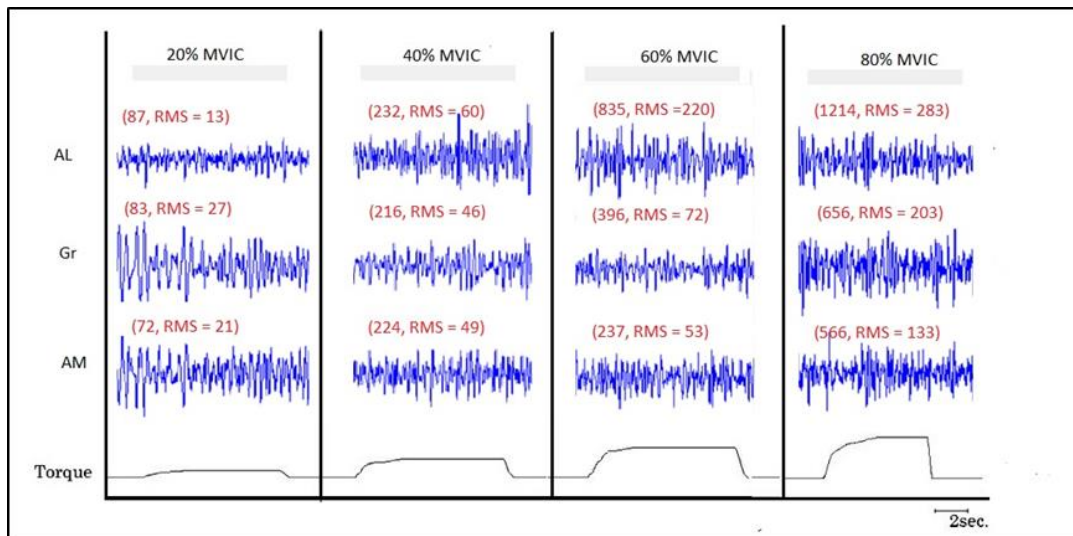
Figure 3-17: An example of one typical participant who participated in the current study. The data represents EMG signals recorded from three muscle components of the adductor muscle group during the ramped protocol of the adductor isometric contraction at 20%, 40%, 60%, and 80% of the MVC. AL: adductor longus; AM: adductor magnus; and Gr: gracilis.
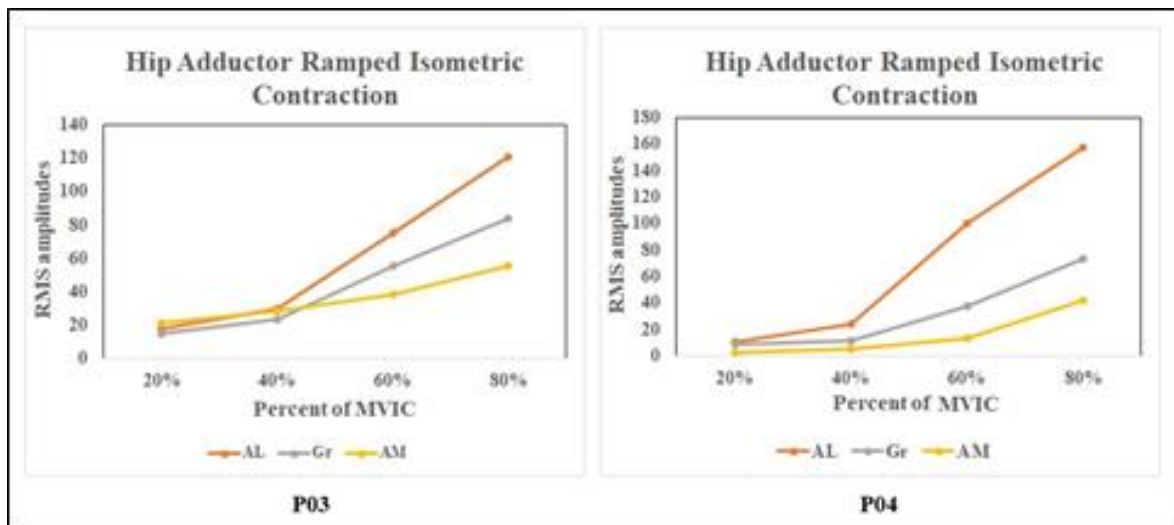
Figure 3-18: An example of two participants who participated in the study. The data represents EMG signals recorded from three muscle components of the adductor muscle group during the ramped protocol of the adductor isometric contraction at 20%, 40%, 60%, and 80% of the MVC. AL: adductor longus; AM: adductor magnus; and Gr: gracilis.

3.6 Discussion
3.6.1 Discussion for Experiment 1

Appropriate electrode placement is critical for gaining reliable information from the surface EMG signal and reducing sources of variability. During walking and running, there is a relatively large amount of hip flexion/extension and therefore there is a potential for relative movement between the muscle and surface of the skin. Consequently, the first experiment of this study aimed to quantify the relative movement of the adductor muscles relative to overlying skin while assuming various hip joint angles (0°, 20°, 40° flexion, 20° extension) in a standing position.

Gait studies illustrated associated hip flexion and extension ranges of $15°$-$35°$ in walking and from $5°$-$50°$ in running (Novacheck, 1998). Such a large range of motion could lead to relatively large changes in the relative position of the electrodes in relation to its corresponding muscle. In the current study, this movement was quantified and it was shown that the surface EMG electrode remains within its corresponding muscle boundary throughout the range of motion. This will minimise the possibly of cross talk from the adjacent muscles. Nevertheless, a number of other factors can also influence the quality of EMG signals for the adductor muscle. These may include the ground impact and subsequent oscillatory movement. Although that these could also affect the validity of EMG signals, these were deemed beyond the scope of this investigation.

The results of this experiment quantified the minimum, maximum, and the average distances between the edge of the muscle and the edge (vertical projection) of the electrode boundary for each of the adductor muscles at various hip flexion/extension joint motion. The results displayed at least 5 mm from each side of the electrode borders for the three adductor muscles throughout the different hip extension/flexion movement. Such distances indicate the

possible dimensions away from the neighbouring muscles. This confirms that the location of the muscle remains within the electrode detection volume. The electrode detection volume is defined as electrical activity picked up from a spherical volume of muscle tissue having a radius equal to the interelectrode distance (Lynn, Bettles, Hughes, & Johnson, 1978). As the surface electrodes were always placed in a longitudinal orientation along the muscle fibres, there is no chance to get signals from the neighbouring muscles.

The total distances from the right and left sides plus the electrode dimension (10 mm) gives the muscle width. The results of this study showed changes in the average distances that did not follow the same behaviour across different muscles at the different joint angles. This finding is consistent with Delaney et al. (2010), who found that at both joint angles (knee flexion at 90° and extension) similar changes in muscle width were produced. They found that changing the knee joint angle from 90° flexion to extension resulted in a 7% reduction in RF width at rest. This slightly smaller potential for change in width during contraction in extension did not appear to have an effect during MVIC, as the change was similar to that in flexion. This support our finding as the change in the joint angle (hip flexion /extension angle) will not affect the relative of the muscle in relation to the overlying skin

Other studies have examined muscle dimensions (e.g. width) changes with different joint angles. For example, Enoka (2015) stated that as a result of changing the joint angle, the muscles contract and move relative to the skin. The actions of contractile elements cause the muscle to be displaced in a transverse direction (across the muscle) relative to the skin. Consequently, this displacement results in an increase in the transverse width of the muscle. Thus, any significant difference in dimensions may affect the relative electrode location and therefore collection volume. The results of our experiment demonstrated minimal changes in muscle distance around the electrode position among different hip flexion/extension

movements that are associated with walking and running. This finding is similar to that showed by Delaney et al. (2010), who examined the contractile ability of the quadriceps muscle using ultrasound imaging. They found that at maximal voluntary contractions the change in RF width was similar in knee extension and 90º flexion.

The possible cause for the similar (right and left) distances among different joint angles observed in the present experiment is changing in angle of pennation rather than muscle width during different joint angles. It was noted that the angle of pennation is strongly related to the joint angle. Such justification is supported by Herbert and Gandevia (1995), who evaluated the changes in pennation of the brachialis muscle with joint angle and muscle torque. They found a robust relationship between elbow angle and brachialis pennation. In addition, they stated that during a maximal isometric contraction, as muscle tension increases, the width remains constant and the pennation angle increases.

3.6.2 Discussion for Experiment 2

The objective of the second experiment was to quantify the relative movement of the adductor muscles with respect to the overlying skin while performing a ramped isometric contraction. The results demonstrated a distance of at least 5 mm from each side of the electrode borders to the muscle boundary for the AL during the ramped protocol of isometric contraction. Although this distance was smaller, approximately 3 mm, in AM and Gr muscles. This was only observed only during the high level of MVIC). Therefore, in line with the findings of the first experiment, changes in isometric muscle contraction did not appear to affect the location of the muscle under the electrode. This demonstrates that isometric muscle contractions that occur during walking or running are unlikely to have a major impact on the validity of EMG signals.

There are no previous studies have looked at changes in muscle position relative to a fixed position on the skin. However, there has been some interest in muscle geometry changes. Therefore, this section is going to contrast the finding of experiment 2 with these studies. The total distances from the right and left sides plus the electrode dimension (10 mm) gives the muscle width during different level of MVIC. The results of the current experiment established that the changes in the average distances did not follow the same behaviour across the different isometric contractions for all adductor muscles. Different behaviour was recorded by Delaney et al. (2010), who found a decrease in RF width. They showed a significant negative correlation with increased forces, and a clear reduction in RF width from rest 30% MVIC but no further change at 75% MVIC.

In the current study, the results found demonstrates relatively constant right and left distances (which is part of the muscle width) among different levels of isometric contraction. This indicates that the electrodes remained at the same place over the tested muscles during the various level of MVIC. The findings of this study are similar to those found by Delaney et al. (2010), who examined the contractile capability of the quadriceps muscle using ultrasound imaging during submaximal and maximal voluntary contractions. They found significant differences in RF width (total distance between muscle borders) between different contraction levels, except between 20% to 30% MVIC, and 75% to MVIC (similar to the present findings). Testing a different muscle and using a different incremental protocol could be the cause of the different findings. On the contrary, Rainoldi et al. (2000) who assessed the geometrical artefacts on surface EMG during isometric contractions (0, 50, and 70% MVIC), found that the muscle slides relative to the overlying skin. They demonstrated a 1 cm shift for the VMO muscle in three out of five subjects. For the VLO muscle, a 1 cm shift was noticed in two out of four participants. Such a shift produces the main contribution to geometrical

artefacts. The sample number, the variation among the isometric contraction levels, knee joint angles, and the different tested muscles may have caused the difference.

Possible explanations for the conflicting findings between this and previous studies (Delaney et al., 2010; Rainoldi et al., 2000) include the different %MVIC used to activate the muscles as well as the difference in pennation angles. The large and powerful AM is more pennated; its angle of pennation is around 16 degrees. The adductor brevis, AL, and Gr are less pennated, with angles of around 6-8 degrees (Ward et al., 2009), whereas RF is not pennate (Delaney et al., 2010). The explanation concerning the difference in the pennation angle is supported by Gans and Bock (1965) and Gans (1982), who stated that during contraction, thickening of pennately arranged fibres is compensated by changing the fibre angle as they shorten. Accordingly, surfaces of origin and insertion remain parallel and equidistant. Hence, the skin movement with underlying muscles did not change during different levels of isometric contraction. Subsequently, the estimated electrode placement did not change with different %MVIC and also pick up EMG signal from the muscle of interest.

3.6.3 Discussion on experiment 3

The purpose of this experiment was to investigate the EMG-force relationship of the adductor muscles during ramped isometric hip adductor torque. To the best of our knowledge, this is the first attempt to explore the EMG-torque relationship of the adductor muscles using surface EMG electrodes. This relationship was confirmed by the regression model where, increasing isometric torque by 1% increased the AL, AM and Gr activity by 4.0, 3.3 and 3.8%, respectively. This could be interpreted as that the isometric torque is potentially important predictor of the muscle activity. The reason behind this relation is that both EMG activity of the muscles and their corresponding force depends on the number of motor units and their firing rates (Merletti and Parker, 2004). During contraction, the extent to which the

motor units are activated determines the amount of tensile force produced (Schoenfeld, 2010).

The results of the current study revealed that there is a clear force-EMG relationship under isometric conditions. However, it was appreciated that this study did not address the relation between dynamic contractions and EMG amplitudes, it is recommended that this dynamic relationship will still be monotonic as any kind of EMG muscle contraction relation will rely on the number of motor units and their firing rates. Therefore, regardless the type of contraction, a relationship should exist between the EMG amplitudes and the corresponding force, although the nature of relation could differ. Therefore, the definite linear relationship between EMG amplitude and the isometric torque, demonstrated in the current study, provides confidence that the proposed protocol will afford a useful indication of the level of activation produced by the adductor muscles.

The findings of this study are similar to those found by Perry and Bekey (1981); Lawrence and De Luca (1983); Woods and Bigland-Ritchie (1983); Alkner et al. (2000) who proposed the existence of a relatively close relationship between muscle force and EMG activity for the biceps brachii, deltoid, soleus and quadriceps femoris muscles under isometric conditions. In addition, this finding is consistent with those of Bilodeau, Schindler-Ivens, Williams, Chandran, and Sharma (2003) who revealed positive correlation between the EMG RMS amplitude for the RF, VMO, and VLO and the force in men as opposed to in women. In the same manner, Gerdle, Henriksson-Larsen, Lorentzon, and Wretling (1991) reported a highly significant positive relationship between torque and signal amplitude in the investigated muscles (RF, VL, and VM) in a group of healthy females.

3.7 Limitations

The current study is limited to specific cohort (lean male subjects). This choice was made in order to decrease the subcutaneous fat layer to its minimum level, thus, decreasing the effect of fat layer on EMG signals to it minimal level. In the current study, it was not possible to monitor the oscillatory movement of the thigh during running. Although this could have an impact of the relative position of the muscle and the electrode, this phenomenon was not investigated in the current study and so further research is needed. This study is also limited as it only investigated the relation between force and EMG amplitudes during isometric contraction. Further studies are needed to explore the relation during the other types of muscle contraction.

3.8 Conclusion

Despite the importance of the adductor muscles, there is a paucity of studies investigating the activation patterns of these muscles during gait. A possible reason behind this lack of research is the difficulty in measuring the activity of these muscles. Therefore, this study aimed to understand the potential for using surface EMG to measure activity in this muscle group. In order to achieve this aim, three separate but complementary experiments were conducted. In the first experiment, the relative movement of the adductor muscles under the skin was quantified at different hip joint angles. In the second experiment, the positions of the adductor muscles were quantified during a ramped isometric contraction in order to identify the effect of muscle contraction on the relative position of the muscle. In the third experiment, the relationship between adductor torque and the magnitude of the adductor EMG signals during ramped isometric contraction was explored.

The most important finding of this study is that the electrodes remain at over the tested muscles during different hip extension and flexion movements and during the various levels

of muscle contraction. In addition, a linear relation was found between the torque and EMG amplitude using data collected from EMG electrodes placed over the adductor muscles using the proposed protocol. Together these findings demonstrated that EMG data collected using the proposed protocol is unlikely to be affected by cross talk and that the amplitude of the signals is likely to reflect the level of force produced during dynamic movement.

Chapter 4: Consistency of EMG and kinematic variables in walking and running

In the previous chapter, a standard method was developed to locate and measure the EMG activity for the adductor muscles. Further work is now required to ascertain whether this method is reliable across different testing sessions. Therefore, this chapter aims to explore the degree of consistency between EMG measurements (from lower limb muscles) collected during gait on two different occasions. A further complementary aim is to recognise the most proper normalisation approach for EMG processing for this group of muscles. The experimental procedures described in Chapter 3, which were carried out to locate the adductor muscle using the ultrasound, were applied in the present study. The paucity of reliability EMG studies during running inspired us to add the majority of the lower limb muscles to the testing protocol for this study. This will provide a comprehensive study of the repeatability of the electromyographic activity of lower limb muscles during running.

4.1 Background

In this section, the importance of reliability measurement for interpreting EMG activity during the dynamic tasks is first discussed. This is followed by a discussed of the sources of variability in EMG measurements. The variability may occur as result of intrinsic or extrinsic factors. The methods used to minimise the effect of these factors in EMG measurements are then discussed. Finally, the previous studies that discussed the reliability of EMG measurement in walking and running are summarised.

4.1.1 Reliability and its importance

Reliability is the extent to which measurements under same testing conditions are stable, consistent, and dependable (Portney, 2009). In addition, it denotes to the stability and

consistency of measures in relation to time, so that differences between measures result from variations in the variable being measured (Keskula, Dowling, Davis, Finley, & Dell'Omo, 1995). For clinical implementations, the consistency of the EMG measurements is essential to diagnose and categorise neuromuscular disorders (Gavilanes, Goiriena, & Tobar, 2000). If critical clinical choices are to be made based on representative data from this single evaluation, it is essential to inspect the reliability of gait data on a cycle-to-cycle, run-to-run, and day to-day basis (Kadaba et al., 1985). A number of instrumentations, experimental protocols, and processing techniques have been used to assess the repeatability of EMG patterns during locomotion (Kadaba et al., 1985; Kleissen, Litjens, Baten, Harlaar, & Hof, 1997; Winter & Yack, 1987). However, a small number of studies have been determined the reliability across different testing sessions.

4.1.2 Sources of variability in EMG

A number of factors could influence the measurement of the muscle activity in clinical or laboratory settings. These factors include extrinsic and intrinsic factors (Itoh, Kimura, & Wakayama, 2016). It is important to minimise the variation of EMG signals caused by intrinsic and extrinsic factors in order to improve the clinical value of EMG data (Burden & Bartlett, 1999). The extrinsic factors include electrode properties, electrode placement, , humidity, and fluctuations in movement speed (Baur, Hirschmuller, Muller, Gollhofer, & Mayer, 2007). The extrinsic or rater-related factors can be controlled through standardisation of the measurement process. In contrast, the intrinsic factors, which include the type of muscle fibre, muscle diameter, muscle length, and the amount of soft tissue, vary between subjects and cannot be controlled (Basmajian & De Luca, 1985; Burden & Bartlett, 1999). Little is known about the factors that could affect EMG measurement for the adductor muscles. Therefore, prior to any kinesiological study, it is important to establish the reliability of the measurement procedure for this group of muscles.

The electrode placement is the most important factor in the measurement of muscle activity during dynamic tasks. Improper electrode placement could possibly pick up the activity of the muscle fibres that are close to each other but are not the muscle of choice, resulting in cross-talk (Winter, 2009). Therefore, recommendations for capturing surface EMG, published in response to the great inconsistency in data collection/processing methodology in order to permit data exchange, are now well established (Merletti, Hermens, et al., 2001). The SENIAM project resulted in guidelines for EMG collection and amplitude estimations, including spectral analysis for surface EMG and a set of test signals. However, the adductors group is not included in this project. To date, the degree of consistency for the adductor muscles between measurements is not clear during overground ambulatory tasks.

4.1.3 Repeatability of the EMG measurement for lower limb muscles during walking

As discussed earlier in the literature review chapter, a limited number of studies have determined the repeatability of EMG measurement during walking. Kadaba et al. (1989) found that repeatability within one particular day was slightly better than between test days for the GMax, GMed, AL, VLO, RF, VMO, LHam, MHam, TA, and MGastro muscles. They calculated the coefficient of multiple correlation (CMC) and found a high within-day and between-days reliability (>0.8, 0.7 respectively) for all muscles, although this was lower for the AL (0.7, 0.6 respectively). Lyytinen et al. (2016) demonstrated that the VMO, TA, and biceps femoris exhibited a good between-day repeatability (ICC ranged from 0.77 to 0.84) in level walking. Murley et al. (2010) concluded that the normalisation techniques have an impact on the reliability of EMG parameters for the lower-leg muscle muscles. The TA and MGastro displayed good to excellent reliability when normalised to sub-maximal voluntary contraction compared to normalisation to maximum voluntary contraction.

The reliability of EMG characteristics during walking has also been determined in patients with musculoskeletal conditions. Hubley-Kozey et al. (2013) found good to excellent ICC values (> 0.8) in individuals with moderate knee osteoarthritis (OA) for the MGastro and LGastro, VMO, VLO, RF, and MHam and LHam muscles. Similarly, the EMG profile of the MGastro exhibited excellent between-days repeatability during level walking for OA individuals (Lyytinen et al., 2016). In addition, the CMC values were greater in the shod condition than the barefoot condition in rheumatoid arthritis patients associated with pes planovalgus for TA, soleus, peroneus longus, and MGastro muscles (Barn et al., 2012). However, no study has examined the reliability of EMG measurement for the set of three superficial adductor muscles during walking for either healthy or unhealthy participants.

4.1.4 Repeatability of the EMG measurement for lower limb muscle during running

Only a small number of studies have been carried out to understand the reliability of EMG measurement during running. Importantly, these studies aimed to determine the reliability of EMG parameters during running within a single session and not across different testing sessions. For example, Smoliga et al. (2010) recorded EMG from the legs (VLO, semimembranosus, GMax, and RF), torso, and arm muscles during running. They found good within day-reliability (ICC > 0.80) for the parameters studied (integrated EMG, root mean square EMG, maximum M-wave, and median power frequency). EMG within day-reliability repeatability has also been assessed during treadmill running at different velocities for five lower-extremity muscles (Karamanidis et al., 2004). They found that the reproducibility of the EMG data for the posterior leg muscles was better than for the anterior leg and thigh muscles (i.e. the MGastro and LGastro showed high reliability, while the VLO, hamstrings, and TA demonstrated low reliability).

In the same context, Golhofer et al. (1990) showed good intra-individual stability for the gastrocnemius and soleus during running. The activity of VMO, VLO, RF, biceps femoris, and MGastro appears to be more repeatable when normalised to maximum voluntary contraction or peak running speed (ICC > 0.80) compared to normalisation to 70% peak running speed (Albertus-Kajee et al., 2011). The repeatability of the EMG profile for the adductor muscles during running has not been discussed in the literature. Therefore, there is a need to understand the level of EMG repeatability for this group of muscles during running.

4.1.5 Reliability indices

This section demonstrates the different methods commonly used to measure the reliability of measurement. In addition, the pros and cons of each reliability index are highlighted. Although the researchers have chosen to use a diversity of terms to discuss reliability, from a statistical point of view, the terms reliability, reproducibility, repeatability, and consistency are synonymous. However, the term reproducibility is the term that best reflects the central question of the ability to attain similar results on repeated testing. Moreover, the assessment of both reproducibility and precision require repeated testing, but precision focuses on the magnitude of the measurement reproducibility error (Knutson et al., 1994). A range of parameters have been identified and broadly used to measure variability and reproducibility. The following paragraphs will discuss the most common reliability indices.

The coefficient of variance (CV) is the standard deviation (SD) divided by its mean value. It is not indicative of reproducibility but more the precision of measurement (Knutson et al., 1994). It can be computed between-individuals (LeVeau & Andersson, 1992; Limbird, Shiavi, Frazer, & Borra, 1988) or within-individual depending on the purpose of the study (Winter, 1984). Moreover, it can be averaged across the gait cycle to indicate the overall variability for any gait variables. In addition, it depicts the distribution of data around the mean.

The within-subject CV is more related to reproducibility because it depends on subject-repeated measures, thus estimating the value of pure measurement error. Conversely, the between-sessions CV can be considered as reliability index because it is the estimation of pure measurement error within, and the between-individuals CV accounts for the variation between subjects (Knutson et al., 1994). Nevertheless, there may be a possible problem when the SD is equal in two different cohorts with different mean values. The CV in data with a low mean value can be higher, which could lead to misinterpretation.

The variance ratio (VR) was firstly proposed by Hershler and Milner (1978), and it can be estimated as the sum of the variance at each time point within a gait cycle divided by the total variance of the data. The VR is used to assess the repeatability of both normalised and non-normalised gait EMGs (Burden et al., 2003; Kadaba et al., 1985). In addition, this measure of variability is independent of the number of strides analysed (Gabel & Brand, 1994). Burden et al. (2003) used the CV and VR as between-session variability to compare the normalised and non-normalised profiles. However, similar to the CV, the VR is a ratio to reference value and does not reflect the absolute nature of the signal (microvolt or percentage of normalised EMG). This more likely misleads the interpretation of the data, as when the actual value is small, the ratio will be large. Moreover, without any information on the actual signal, the clinical interpretation of this ratio can be challenging.

The intra-class correlation coefficient (ICC) was originated by Fisher (1958) and expresses reliability as the ratio of true score variance to total variance, including the potential error components (Wastell & Barker, 1988). Recently, the ICC has been calculated by mean squares (i.e. estimates of the population variances based on the variability among a given set of measures) obtained through analysis of variance (Koo & Li, 2016). It is used to compute the variance of a single measurement at multiple time points across the gait cycle and is

averaged, as with the CV (Francis, 1986). There are several kind of forms of the ICC that can give various results when applied to the same set of data and consequently provide different interpretations. Shrout and Fleiss (1979) demonstrated six forms of the ICC for a reliability study with a selection guideline. Portney (2009) suggested the values for acceptable reliability using the ICC ranges from 0.00 to 1.00. ICC values less than 0.5 are indicative of poor reliability, values between 0.5 and 0.75 indicate moderate reliability, values between 0.75 and 0.9 indicate good reliability, and values greater than 0.90 indicate excellent reliability. Although researchers should defend their judgments within the context of the specific scores being assessed, there are no standard values for adequate reliability.

Similar to the aforementioned measures, the ICC does not afford information about the actual signal to allow direct interpretation in practice. In addition, any systematic changes in measurement could affect the association between the measured values (Knutson et al., 1994). Moreover, the formula for calculating the ICC shows that ICC measurement depends on the total variance of the population and not just measurement variability. Thus, the ICC should be conducted on samples with the same characteristics to avoid misinterpretation. Accordingly, a low ICC could not only reflect the low degree of rater or measurement agreement but also relate to the lack of variability among the sampled subjects, the small number of subjects, and the small number of raters being tested (Lee et al., 2012; Portney, 2009). Therefore, there is no true gold standard for the measures of reliability or criteria suggesting the clinical meaning of these values.

The coefficient of multiple correlation (CMC) is another repeatability measure calculated from the positive square root of the adjusted coefficient of the multiple determination. It is calculated for a true estimate of within-day and between-days repeatability (Kadaba et al., 1989). It measures the overall similarity of waveforms, taking into account the concurrent

effects of differences in offset, correlation, and gain (Ferrari, Cutti, & Cappello, 2010). For example, in the inter-rater reliability study, CMC values increased when offset was removed, which indicated a high level of similarity between gait curves (Røislien, Skare, Opheim, & Rennie, 2012). Although the CMC itself depends on data collection, CMC methods represent only a fraction of the reliability of gait curve data. Itoh et al. (2016) showed a conflict between methods using the ICC and CMC, as some CMC-based data showed high or low reliability, whereas ICC-based data showed the contrary. Since one of the aims in this thesis is to describe the profile of the lower limb muscles, CMC could be the most appropriate method for determining the similarity of the wave form. Additionally, in order to improve the assessment of the consistency and precision between different measurement occasions, standard error of measurement has to be calculated.

The standard error of measurement (SEM) is defined as the difference between measurements of the same quantity on the same subject (Bland & Altman, 1996). It is calculated by multiplying the standard deviation of the measurements by the square root of one minus the reliability coefficient (Keskula et al., 1995). The estimation of the SEM is based on repeated measurements from a single individual or a set of scores obtained from a larger sample of subjects, where these scores can be anticipated on retesting depend on a confidence interval (CI). Therefore, it is one of the most communal statistical methods used to express response constancy by calculating the standard error in a group of repeated scores and establishing reliability (Portney, 2009). Unlike the ICC, the SEM quantifies error in the same measuring unit such as degree or microvolt and provides the opportunity to calculate the range where the subject's true score is located (Stratford & Goldsmith, 1997). Thus, it has a clinical meaning, allowing a clinical interpretation.

The reliability indices presented above demonstrate that there is no gold-standard statistical method for measuring the consistency of measurements between testing days. Each method has its own advantages and disadvantages. Therefore, it is important to use at least two different methods, including the SEM, in order to have clinically meaningful results in reliability studies.

4.1.6 Choosing the appropriate normalisation technique

As previously mentioned in the literature review (section 2.8.2), several techniques are widely used to normalise the EMG amplitude. The best normalisation must lead to high reliability and low variability between testing occasions. The normalisation should not rise the natural variability between subjects but diminish the systematic or measurement variability to permit the detection of any pathological sign. For gait analysis, the normalisation can be categorised into self-normalisation, which relates to a criterion inside the signal being recorded such as peak/ mean of the dynamic amplitude of the gait cycles, and external normalisation, which relates to a criterion of a separate measure such as MVIC. Other external normalisations such as the submaximal isometric voluntary contraction (Isometric-subMVIC), submaximal dynamic voluntary contraction (Dynamic-subMVIC), and angle and velocity specific maximal isokinetic voluntary *(*Isokinetic-specMVIC*)* can be used. However, the most common normalisation method in gait studies is MVIC (Dubo et al., 1976; Fuglevand et al., 1992; Hermens et al., 1999).

The influence of choosing MVIC, peak, and mean on the reliability of EMG from lower limb muscles is still indecisive. For knee extensors and flexors, MVIC has been shown to improve the between session reliability of the raw EMG, but the reliability is still lower than the peak and mean normalisations (Burden et al., 2003). In another study, it was found that MVIC resulted in the greatest repeatability within day performance compared to a mean and a peak

in gastrocnemius during walking (Knutson et al., 1994). Murley et al. (2010) concluded that normalisation to peak activity obtained from fastest walking pace resulted in higher within and between-session reliability than MVIC normalisation in most of the studied muscles. This may propose that MVIC as a method of normalisation is influenced by the tested muscles and the type of reliability study (within- or between-days reliability study). Although peak and mean normalisation methods result in the most homogenous ensemble averaged profiles, the true variations within a group could be removed (Burden et al., 2003). This may result in a false positive clinical interpretation (Knutson et al., 1994). In addition, the normalised amplitude to the mean/peak of the dynamic task does not imply the desired level of muscular activity during gait but the relative activity to the mean or peak in various phases.

Many studies have compared the different normalisation techniques used for superficial muscles. MVIC has great potential for clinical application, as the normalised EMG would be presented as a percentage of the maximal activation capacity of the muscle, while the mean and peak would be presented as a percentage of the EMG relative to the reference values. However, there is no study investigating the effect of normalisation for the adductor muscles during walking and running tasks.

4.1.7 Sources of variability in kinematic variables

Two major factors could influence the reproducibility of kinematic measurements in gait analysis: the variability in human performance and variability in the measurement process (e.g. instrumental errors, skin movement artefacts, kinematic model assumptions) (Growney, Meglan, Johnson, Cahalan, & An, 1997; Hopkins, 2000). The error in the measurement process may include: the accuracy of hardware and software, the reliability of the marker placement across sessions (test-retest) and across testers (inter-tester), skin-marker movement

artefact, and the testing protocol (Tsushima, Morris, & McGinley, 2003). Human performance, the other source of variability, may arise as a result of recruiting diverse of ages and genders and the motor control strategies used in performance (Growney et al., 1997). Having proper inclusion and exclusion criteria and an accurate testing procedure could minimise the effect of these factors on the consistency of kinematic measurements between testing days.

4.1.8 Repeatability of kinematic measurements in walking

Researchers quantify the reliability of kinematic variables in walking in a variety of ways, using between-subject and within-subject variations. The latter is considered as the most important kind of reliability measure for researchers, because it influences the precision of estimates of variation in the variable of an experimental study (Hopkins, 2000). Moreover, the degree of joint angle has been measured in different planes during walking both within-day and between-days. It has been proposed that motion in the sagittal plane, the frontal plane, and the transverse plane, excluding pelvic tilt, is highly repeatable in both test-retest and inter-tester reliability (Tsushima et al., 2003). In the sagittal plane, intra-subject repeatability is excellent for kinematic data both within a test day and between testing days (Growney et al., 1997; Kadaba et al., 1989; Tsushima et al., 2003). Compared to the sagittal plane, the repeatability for frontal and transverse plane motion is lower during walking. This could happen due to the small range of movement available in these planes as well as the lack of a precise displacement pattern (Growney et al., 1997; Kadaba et al., 1989). Therefore, an accurate marker placement could overcome this issue and result in a highly repeatable dataset.

4.1.9 Repeatability of kinematic measurements in running

Similar to walking, the within-day reliability of the kinematic parameters measured during running is more reliable than between-day reliability. In the same context, different angular kinematic parameters for the hip, knee, ankle, and pelvis display higher reliability in the sagittal plane compared to a relatively lower value in the frontal plane during running tasks (Ferber, Davis, Williams, & Laughton, 2002; Mason, Preece, Bramah, & Herrington, 2016; Noehren, Manal, & Davis, 2010). The difference between planes may be attributed to higher speed (McGinley, Baker, Wolfe, & Morris, 2009). This increase in speed could lead to an increase in soft tissue movement, especially in the skin-mounted marker sets. In addition, the natural degree of variability between trials could be another source of variability between testing days. All these reasons for variability will be combined with the instrumental errors of a motion capture system (ideally 1°) and any day-to-day inaccuracy of marker placement to give a fundamental limit to the accuracy of kinematic measurement.

To overcome the aforementioned difficulties in developing repeatable EMG data and in choosing the best method for normalising EMG amplitudes, the present study was carried out. Additionally, in order to compare with previous running studies, the kinematic reproducibility of lower-limb kinematics is quantified. This chapter aims to answer the following research questions:

- Do healthy subjects demonstrate consistent EMG patterns during walking and running in repeated sessions, separated by approximately 1 week?
- Do the different normalisation methods affect the reliability of the lower muscles in walking and running gait?
- Do healthy subjects demonstrate consistent kinematic patterns during walking and running in repeated sessions, separated by approximately 1 week?

4.2 Aims

The first aim of this study was to investigate the degree of consistency between EMG measurements (from lower limb muscles) collected during overground walking and running on two different occasions. The second complementary aim was to compare the different normalisation methods of muscle activation during walking and running from the relevant methods available within the literature. The final aim was to assess the between-day reliability of measuring 3D biomechanical variables during walking and running activities.

4.3 Methodology

This section provides an overview of the procedures for the current study. These procedures included the recruitment methods, inclusion and exclusion criteria, and the equipment used in the study. Several pieces of equipment were utilised in this study, such as ultrasound imaging, a direct transmission system, and a motion-analysis system. In addition, the method of application for the surface EMG measurement and the 3D marker placement is described. Finally, the testing protocol during the dynamic task and during measuring the maximum isometric contraction will be explained.

4.3.1 Recruitment plan

A number of avenues for recruiting the participants were used for this study. Firstly, posters were placed around the university campus. Secondly, invitation emails were sent to local running/triathlon clubs and clients who had previously used the commercial running performance clinic at the University of Salford. Finally, an advert was displayed on the running performance clinic website. Many responses were received, and only individuals who met the entry criteria, outlined below, were invited to participate in this study.

4.3.2 The entry criteria

a. All subjects were males and aged between 18 and 40 years of age. This age range was carefully chosen to represent the young, athletic population to whom the outcomes of the study are most likely to be implemented.

b. Subjects needed to be free from lower limb insult for a minimum of six months before the testing and have no history of lower limb surgery. Injury was defined as any musculoskeletal disorder that prevent the subject from performing his normal exercise habit.

c. The regular training programme for those participants needed to include running training at least three times per week, for a minimum total of 10 miles per week. In addition, they had to be carrying out their regular training routines for a minimum of three months before joining the current study.

d. The participant needed to be able to perform the MVIC tasks comfortably.

e. The participant could not have been participating in another injury prevention programme, particularly a programme focused on changing pelvic kinematics, as this may have altered the natural relationship between frontal plane pelvic kinematics and adductor muscle activation.

4.3.3 Participants

Using the aforementioned criteria, a cohort of ten male recreational runners, with no history of lower limb injury or surgery, was recruited for this study. The mean (SD) age of the subjects was 30 (7.2) years, mean (SD) height 1.74 (0.1) m, mean (SD) weight 70.1 (7.5) kg, and the mean (SD) body mass index (BMI) 23.1 (1.4) kg/m^2. With no prior data, it was not possible to afford a formal estimation of the required sample size. Similar studies have been carried out by Kadaba et al. (1985); Luginbuehl et al. (2013); Luginbuehl et al. (2016); Ochia and Cavanagh (2007); Sinclair, Brooks, Edmundson, and Hobbs (2012) using a sample size

of 10 participants to assess the reliability of EMG measurement during both walking and running. Moreover, none of the studies reviewed in the background section did a sample-size calculation for reliability testing. Therefore, a sample of ten healthy participants was chosen for this study. Such a sample size is convenient to explore these technical issues regarding surface EMG and other biomechanical variables.

4.3.4 Ethical approval

Before commencing the first data collection session, all subjects read and signed a written informed consent statement approved by the Research, Innovation, and Academic Engagement Ethical Approval Panel at the University of Salford (appendices V and VI).

4.3.5 Instrumentation

The same equipment used in the previous study was used here in the present study as well. This included an ultrasound imaging system (MyLab70, Esaote, USA), a Direct Transmission System with 16 channels (Noraxon USA inc., model 586 Tele Myo DTS Desk Receiver), the DTS sensors (model 542), EMG lead (542AP), and a disposable adhesive Ag/AgCl EMG electrode (for more details, see the methodology section in Chapter 3). In addition, the set-up procedure for the ultrasound and EMG measurements used in the previous study was followed here (see sections 3.4.3 and 3.4.5). A motion capture system (ten Pro-Reflex, Qualisys cameras with three embedded force platforms) was used in this study in order to identify the gait events and to obtain the kinematic and kinetic data.

4.3.5.1 Three-dimensional system (3D)

A motion-analysis system consisting of ten cameras (Pro-Reflex, Qualisys), with a sample frequency of 250 Hz, and three force platforms (AMTI, USA) embedded into the running track, sampled at 1200 Hz, was used to gather biomechanical data for lower limbs. This

system uses infrared (IR) cameras and passive retro-reflective markers. To enable connection to the cameras, Qualisys proprietary software, Qualisys Track Manager (QTM), was used. There are three stages in the collection of coordinate data using the Qualisys Pro-reflex system: calibration, data collection, and 3D reconstruction of retroreflective markers.

The capture volume size is an essential issue, since it influences the system resolution accordingly, or the accuracy with which position data can be collected (Figure 4- 1). The most proper camera position is that which minifies the blind space surrounding the selected capture volume in the cameras' field of view (Pantano, White, Gilchrist, & Leddy, 2005; Richards, Thewlis, Selfe, Cunningham, & Hayes, 2008). Since the variables of interest in the current study were collected during the stance phase of walking and running, the ten cameras were placed in an umbrella configuration around the three force platforms to assure that they could accommodate the required movements (Figure 4-2). A Brower Timing Gate System (TC-Timing System, USA) was used to monitor walking and running times.
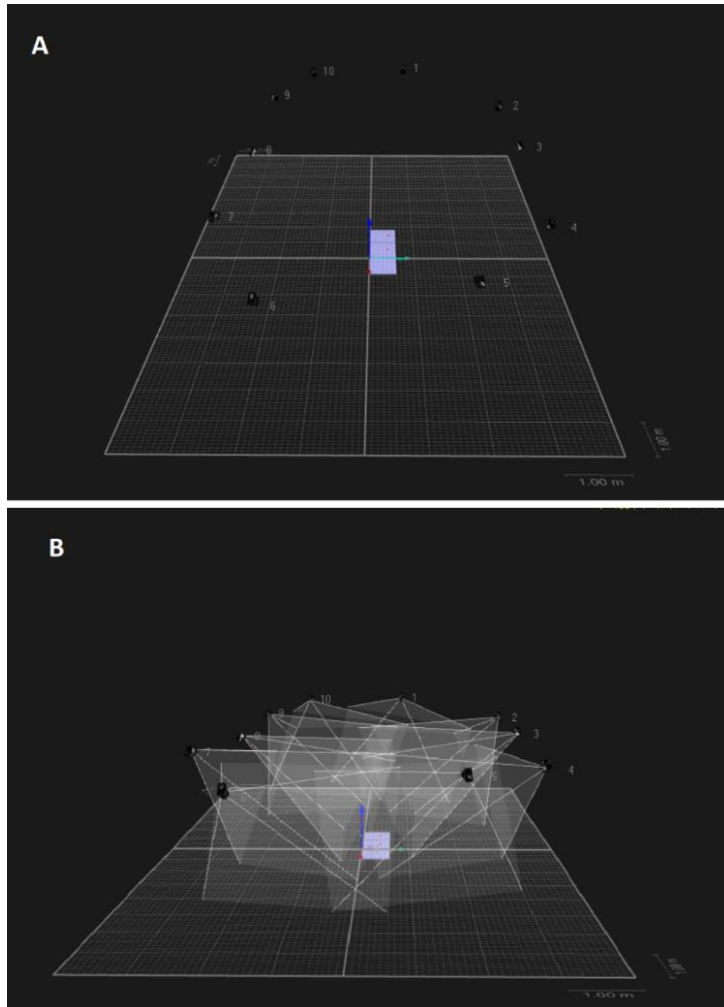
Figure 4- 1: A. Distribution of the camera inside the lab; B. The collection volume of each camera.
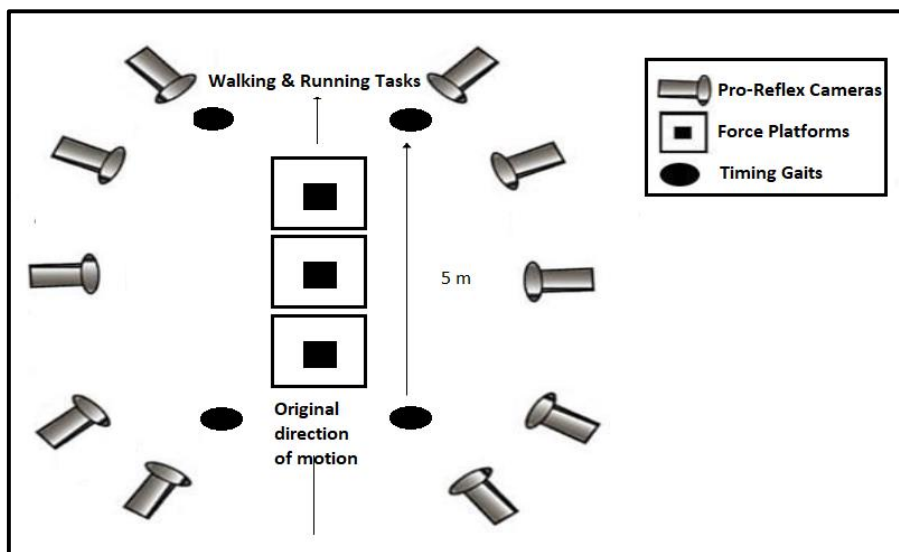
Figure 4-2: The lab configurations.

4.3.5.2 System calibration

Each IR camera affords a 2D image that must be converted into a 3D workplace for the analysis of coordinate data. The reason of doing this is to assure the creation of 3D coordinates of marker positions using a direct-linear transformation method and to facilitate global references (Richards et al., 2008). Marker position in 3D space can only be located according to the accuracy with which the system is calibrated (Payton, Bartlett, British Association of, & Exercise, 2008). The lower the residuals, the more precise the calibration and 3D marker coordinates from measurements.

In order to convert the given 2D image of cameras into a 3D workspace, a static calibration was undertaken. A rigid L-shaped metal frame with four mounted markers, which are situated at known locations and distances to each other, was used in the static calibration of the motion-capture system and to define its relationship to the laboratory reference frame (Figure 4- 3B). This reference L-shaped frame was positioned on the corner of the first force platform and aligned carefully with the two sides of the implanted force plate. A handheld T-shaped wand, with two reflective markers positioned at each end with a fixed and known distance of 601.7 mm (Figure 4- 3A), was used to calibrate the volume that was used during dynamic trials. The calibration process was performed by a random movement of the wand around the test space for 60 seconds. This period, the 60 seconds, was used to enable that volume to be successfully calibrated and to ensure that both lower and higher-floor levels were covered completely (i.e. at least two cameras could see the markers on the wand) (Richards et al., 2008). The T-shaped wand was moved during the calibration time in as many orientations as possible to ensure that the capture volume was covered completely. During the calibration, the position of each marker relative to the origin of the global coordinate system (Lab System) was collected by cameras and recorded in the computer. The L-shaped frame was kept in position on the platform during the calibration (Winter, 2009).
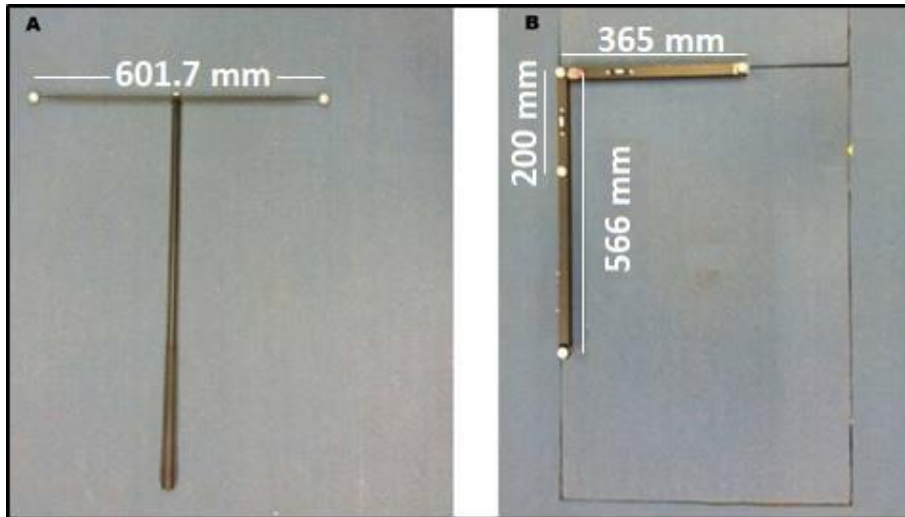
Figure 4- 3: A. Handheld wand and B. Calibration L-frame.

The calibration results inform the user whether the undertaken calibration was successful or not. The resultant values of the calibration process indicate the difference between the factory-measured distance of the static markers on the L-frame (601.7 mm) and the calculated distance based on the actual marker coordinates of the wand in the lab coordinate system. In the current study, the aim was to ensure that the difference between the actual and factory-measured distance of the static markers on the L-frame (referred to as the residual error) was as low as possible. To achieve this, a calibration was only considered successful if the standard deviation of the wand length was below 1 mm for all cameras (Figure 4- 4).
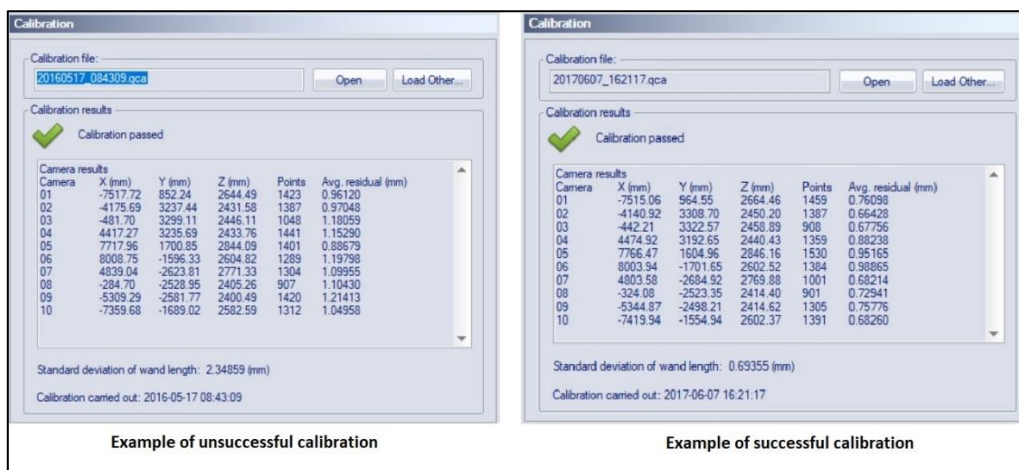
Figure 4- 4: Examples of unsuccessful and successful calibration.

120

4.3.5.3 Coordinate system

The global reference system, the Qualisys Proreflex^{TM} system's Cartesian coordinate system (CCS), was defined using the calibration frame. Each marker position was described with a CCS in which the orientation of the coordinate system was defined. The three axes of the 3D CCS are commonly symbolised as x, y, and z. The system adopted in this study was recommended by the International Society of Biomechanics (ISB) (Wu & Cavanagh, 1995). The ISB suggest that X- axis points in the anteroposterior direction with positive values along the line of progression, while the Y-axis points in vertical directions with positive values in the upward directions. In addition, the Z-axis points in mediolateral directions with positive values with the rightward orientations. This system is based on the rule of thumb for the right hand in which the right thumb is pointed along an axis in positive direction, the fingers then curl around that axis and point in the direction a positive rotation is occurring. The rotation around the z-axis represents flexion/extension movement, while that around the x-axis represents abduction/adduction movement, and the rotation around the y-axis represents internal/external movement (Figure 4- 5).
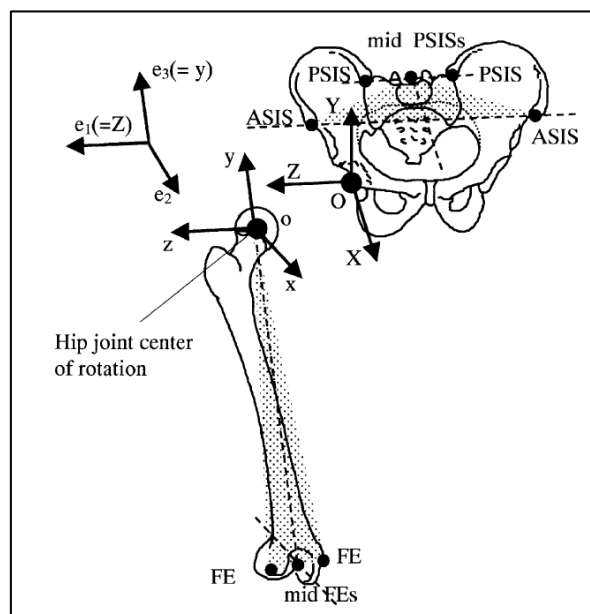
Figure 4- 5: Illustration of the pelvic coordinate system (XYZ), femoral coordinate system (xyz), and the JCS for the right hip joint (Adopted from (Wu et al., 2002).

4.3.5.4 Force platforms

Force platforms measure the body's reaction to the gravitational force and the amount to which the body's dynamic actions vary at the floor level. Three force platforms (AMTI, USA) embedded into the running track, sampled at 1200 Hz, were used in the current study. When a participant walks or runs over the force plates, the forces exerted onto the plates were transmitted from body onto earth. In agreement with the Newton's third law, the plates return the force in an equal and opposite direction back towards the moving subject. When a foot contact on the force plate was available, the gait events (foot strike and toe off) were determined from the vertical ground reaction force based on a threshold set to 20 N, i.e. the initial contact was quantified as the first occasion at which the vertical component of the GRF was greater than 20N, while the toe-off was determined to be the first occasion in which the vertical GRF fell below 20N (Sinclair, Edmundson, Brooks, & Hobbs, 2011).

4.3.6 Surface EMG electrode and 3D marker placement
4.3.6.1 Surface EMG electrode placement

After signing the consent form, the participants were then requested to change into their shorts and a comfortable t-shirt. Firstly, the mass and height of the participants were measured followed by surface EMG electrodes placement. The measured muscles were located according to the SENIAM guidelines (Hermens et al., 2000; Sacco, Gomes, Otuzi, Pripas, & Onodera, 2009). Surface EMG data was obtained from the GMax, GMed, VMO, VLO, LHam and MHam, TA, and MGastro and LGastro and was synchronised with the motion capture system. For locating the adductor muscle group (AL, AM, and Gr), the same approaching technique used in study 1 was also followed here in this study (see Chapter 3, Table 3-2).

The SENIAM guidelines for electrode placement were used in order to minimise the possibility of crosstalk between muscles. For the GMax, the participant lay down on his face and the electrodes were placed at 50% on the line between the second sacral vertebrae and the greater trochanter. For the GMed, the participant lay down on the contralateral side and electrodes were placed at 50% on the line from the iliac crest to the greater trochanter. For the VMO, the participant sat with the knees in slight flexion and the electrode was placed distally at 80% on the line between the anterior superior iliac spine and the joint space in front of the anterior border of the medial collateral ligament. In order to be parallel to the muscle fibres, the electrode was placed at an angle of 55° to the vertical line. For the VLO, the electrode was placed distally at 2/3 on the line from the anterior superior iliac spine to the lateral side of the patella.
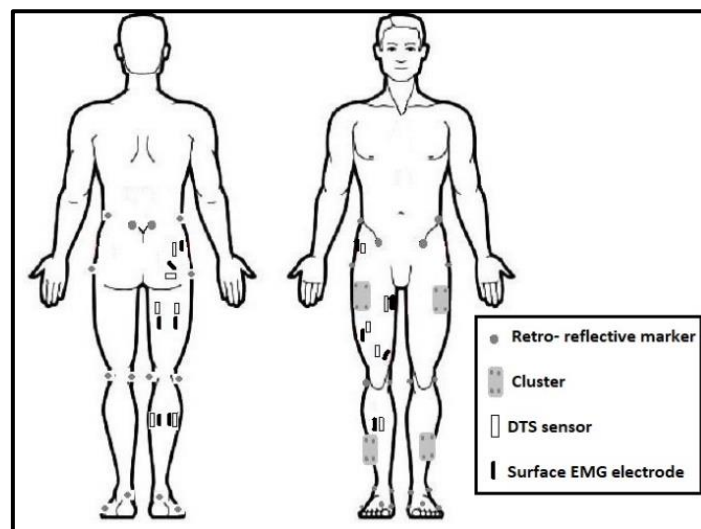
Figure 4- 6: Markers and electrode placement.

For the MHam muscle, the participant lay down on his belly, face down, and the electrode was placed at 50% on the line between the ischial tuberosity and the medial epicondyle of the tibia. For the LHam muscle, the participant again lay prone and the electrode was placed at 50% on the line between the ischial tuberosity and the lateral epicondyle of the tibia. For the TA, the participant lay on his back, and the electrodes needed to be placed at proximal 1/3 on

the line between the tip of the fibula and the tip of the medial malleolus. For the gastrocnemius muscle, the participant lay on his belly, face down, with the knee extended and the foot projecting over the end of the table. For the MGastro, the electrode was placed on the most prominent bulge of the muscle. For the LGastro, the electrode was placed at the proximal 1/3 of the line between the head of the fibula and the heel. Finally, the skin preparation and signal checking process followed in Chapter 3 were followed in this study as well (see Chapter 3, section 3.4.5).

4.3.6.2 Marker placement

Reflective markers of 14.5 mm diameter were used in all trials of data collection. The markers were placed over the skin using hypoallergenic double-adhesive tape attached to a flat-based marker. To define the orientation and position of a segment in three-dimensional space, three non-co-linear markers (cluster) were used (Cappozzo, Catani, Leardini, Benedetti, & Della Croce, 1996), and during capture time, at least two cameras could see each marker at any instant (Payton et al., 2008). A set of 20 markers were positioned on the lower limb of a participant on each side in order to define the anatomical reference frame and centres of joint rotation. Markers were positioned on the lateral and medial aspects of joints, on anatomical landmarks, and at the proximal and distal ends of the segment. Specifically, foot markers were positioned on the first, second, and fifth metatarsal heads and calcaneal tubercle over the standard shoes, ankle markers were attached on the medial and lateral malleolus, knee markers were placed over the lateral and medial femoral condyle, thigh markers were attached on greater trochanter, and finally, pelvis markers were placed over the right and left anterior superior iliac spine (ASIS), right and left posterior superior iliac spine (PSIS), and right and left iliac crest (Figure 4- 6).

Following an acceptable capturing of all the static markers, 12 markers were removed, as they were only needed during static capturing. The remaining 28 markers (16 markers over four cluster plates, eight markers attached to standard shoes, and four markers on ASISs & PSISs) stayed in their position, as they were needed during the dynamic trials. The cluster plates were firmly secured to the antero-lateral aspect of the thigh and shanks of both sides. Manal, McClay, Stanhope, Richards, and Galinat (2000) found that the use of rigid clusters is the optimal configuration compared to individual skin markers (Manal et al., 2000).

4.3.7 Testing protocol
4.3.7.1 Static trials

Before recording the walking and running trials, all subjects were instructed to wear the same shoes (New Balance M539SR, UK) (Figure 4- 7). The reason of doing this was to control the shoe-surface interface and to negate any potential effect on lower limb biomechanics. Thereafter, they started with three minutes of low-intensity warm-up on the walkway at a self-selected speed. Thereafter, they were familiarised with the testing procedure by practising each of the tasks until they felt comfortable with them. After familiarisation, the participant was requested to stand in a stationary position on the force plate in order to capture the static standing trials. The arms of the participant were held away from the waist and thigh in order to clearly see the pelvis and thigh markers. On finishing the above procedures, the participant was ready for recording the dynamic trials.

Figure 4- 7: The standard shoe used during the testing.

4.3.7.2 Dynamic trials

Three-dimensional kinematics of the pelvis, hip, knee, ankle, and ground reaction forces (GRF) were recorded for each participant during the walking trials for ten trials. The participant was asked to walk on a 5 m walkway at 1.25 m.s^{-1} ($\pm$ 0.1 m.s^{-1}), monitored using optical timing gates. A Brower Timing Gate System (TC-Timing System, USA) was used to ensure that each trial was performed within $\pm$ 5% of the selected speed (1.25 m.s^{-1} for walking and 3.2 m.s^{-1} for running). The set of Brower Timing Gate System were set at hip height for all participants. Afterwards, the participant was asked to run on a 5 m walkway for ten trials at 3.2 m.s^{-1} ($\pm$ 0.2 m.s^{-1}). After each trial, the participant walked back slowly to ensure sufficient recovery and to limit the effect of fatigue. All the walking and running trials were recorded during the overground performance. A successful trial required an occurrence of the stance phase on the force plate (AMTI), without an overlap of the foot between force plate and ground floor, within the field of the view of the high-speed motion analysis camera system. In addition, the participant ran or walked within the acceptable range of speed and acceleration ($\pm$ 5% for speed and $\pm$ 10% for acceleration). Unsuccessful trials were ones

whereby less than three markers per segment were visible, speed and acceleration change out of the acceptable range, or a partial/double contact with the force platforms occurred.

Having a single standard speed during running is essential when comparing kinematics and kinetics between and within subjects. This is because changes in speed have been shown to influence the kinematics, kinetics, and the electromyographic activity of the lower extremities. Therefore, most researchers agree that the running speed should be standardised (Colby et al., 2000; Kadaba et al., 1989; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Pollard, Davis, & Hamill, 2004; Queen, Gross, & Liu, 2006). Standardising the speed among participants allows for more accurate interpretation of the results (Stergiou, Bates, & James, 1999). The acceptable overground running speed controlled by photocells in running studies has ranged from 1.5 to 6 m.s^{-1} (Diss, 2001; Ferber et al., 2002; Ferber, Davis, & Williams Iii, 2003; Stergiou et al., 1999; Wank, Frick, & Schmidtbleicher, 1998). The participants were instructed to run at 3.2 m.s^{-1}, as it is a typical running speed for recreational runners and has been used in previous research on healthy runners (Novacheck, 1998) and for runners with running related injuries (Boyer & Derrick, 2015; Burnet & Pidcoe, 2009; Foch & Milner, 2014; McCarthy, Fleming, Donne, & Blanksby, 2015).

Another critical factor which could affect the degree of muscle activation is any acceleration of the subject while performing the individual trials. Newton's second law of motion states that acceleration is directly proportional to the magnitude of the net force, and inversely proportional to its mass ($a = F/m$) (Frost, Cronin, & Newton, 2008). It is known that changes in acceleration are accompanied by change in EMG activity pattern (Frost, Cronin, & Newton, 2010; Sakamoto & Sinclair, 2012; Tokuda et al., 2016). Therefore, to minimise the possible effect of participant accelerations, the change in speed was carefully monitored, via the anteroposterior ground reaction force (AP GRF). With this approach, trials for which

there was evidence of any acceleration or deceleration were rejected. This was quantified by looking at the difference in the braking (-ve) and acceleration (+ve) portion of the AP GRF. When this difference was more than 10% of the total rectified AP GRF force (i.e. the mean absolute impulse), then a trial was rejected (Figure 4- 8). This was done via a custom Matlab programme which was used during the data collection process.
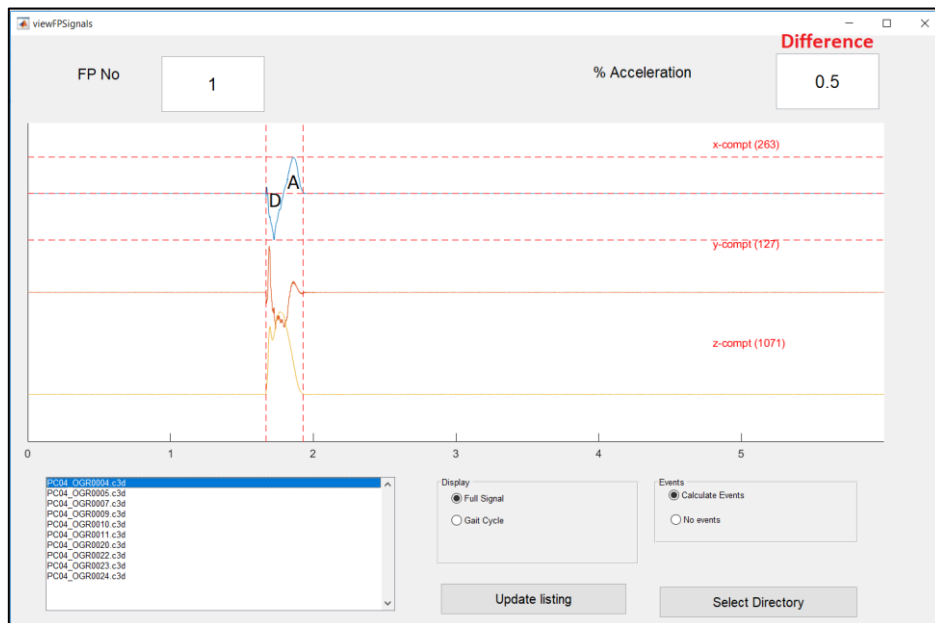
Figure 4- 8: Example of typical accepted trial. The rater looked at the difference between the deceleration phase (D) and the acceleration phase (A) and only trials that were lower than 10% difference were accepted.

4.3.7.3 Maximal voluntary isometric contraction (MVIC)

After finishing the running and walking trials, the participant was asked to perform the MVIC tests for the different muscle groups. Firstly, from prone position with 90˚ knee flexion, the subject was asked to lift his leg up against manual resistance in order to test the GMax muscle (Figure 4- 9A). The same position, lying face down, was also used to assess the maximum contraction of both the MHam and LHam. The participant was asked to bend his tested knee against manual resistance while the hip was in a neutral position and the knee was at 55º flexion (Figure 4- 9B). In the side lying position, the GMed muscle of the top leg was tested by asking the subject to move the tested limb towards the ceiling against manual

resistance at the outer side of the tested knee (Figure 4- 9C). Lying supine, the participant performed two tests for the adductor muscles either from 0º hip flexion (Figure 4- 9D) or from 45º hip flexion (Figure 4- 9E). For both testing positions, the participant was asked to move his tested leg towards his other leg against manual resistance (Lovell et al., 2012).

From the long sitting position, the TA muscle was tested by asking the participant to dorsiflex the ankle and invert the foot against manual resistance without extending his great toe (Figure 4- 9F). From sitting position, the quadriceps muscles (VMO and VL) were tested at 45º of knee flexion by asking the participant to extend his knee against manual resistance (Figure 4- 9G). Finally, the MGastro and LGastro muscles were assessed by asking the participant to stand on his tip toes for his tested limb (Figure 4- 9H). Through all MVIC tests, the participant was verbally encouraged to maximally contract the tested muscle(s) against the manual resistance, hold for three seconds, and then relax. Each test was repeated three times. There was a minute's rest between each contraction to avoid fatigue.

MVIC for the adductors

A pilot study was undertaken to determine the maximum activation levels of the hip adductor muscles in two clinical tests to be used in normalisations of EMG amplitudes obtained from dynamic tasks. As mentioned previously, each participant performed two different tests for the adductor muscles, either from 0º hip flexion (Figure 4- 9D) or from 45º hip flexion (Figure 4- 9E). For both testing positions, the participant was asked to move his tested leg towards his other leg against manual resistance. The participant was verbally encouraged to maximally contract the tested muscle(s) against the manual resistance, hold for three seconds, and then relax. Each test was repeated for three times. There was a 30 seconds rest between each contraction to avoid fatigue.
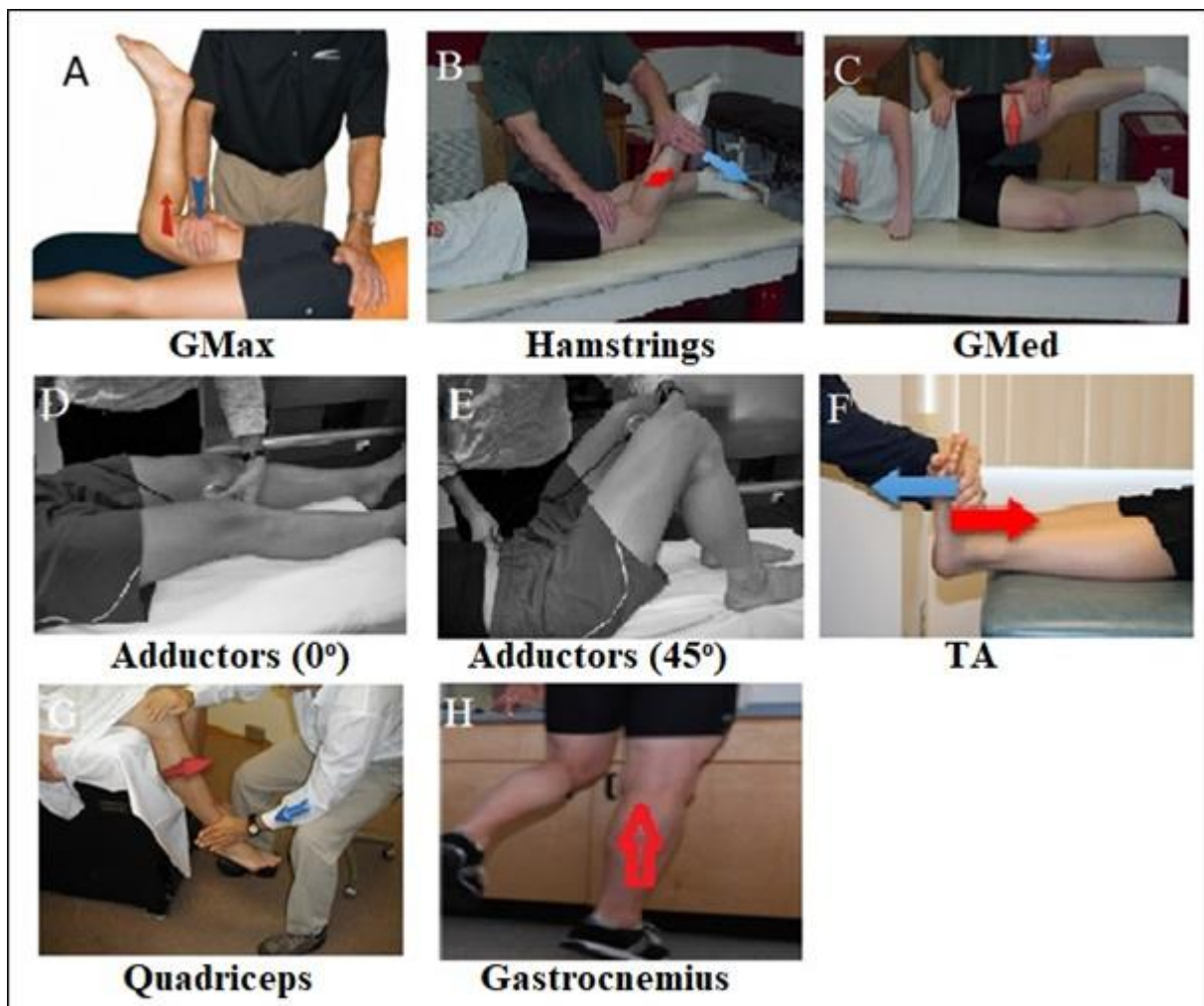
Figure 4- 9: Different positions for testing the MVIC for the lower limb muscle: A. the testing position for the gluteus maximus (GMax), B. The testing position for medial and lateral hamstring muscles at 55⁰ knee flexion, C. The testing position for gluteal medius muscle (GMed), D. the testing position for the hip adductor muscles at 0⁰ hip flexion, E. The testing position for adductor muscles at 45⁰ hip flexion, F. The testing position for tibialis anterior muscle (TA), G. The testing position for quadriceps muscles (vastus medialis obliques and vastus lateralis obliques) at 45⁰, H. the testing position for medial and lateral gastrocnemius muscles from standing position.

This pilot study involved ten volunteers and the results of the highest MVIC values were constantly achieved at 0° for AL and at 45° for Gr and AM across different participants (see Appendix VI). Therefore, given this preliminary work, the protocol summarised in Table 4- 1 was adopted for all subsequent MVIC measurements for the adductor muscles.

Table 4- 1: Summary of the findings of the pilot study.

	Muscle	The best position for measuring MVIC
1	Adductor longus	At neutral hip position (0°) from supine lying position
2	Gracilis	At 45° hip flexion from supine lying position
3	Adductor magnus	At 45° hip flexion from supine lying position

4.3.8 Order of testing

There was a very small chance that the order of the MVIC tests may be influenced by order. Therefore, we randomised the order of the MVIC tests, described above. However, the MVIC tests were always performed after the gait trials to ensure that the gait trials were not influenced by fatigue.

4.3.9 Testing the repeatability of measurement

In order to test the repeatability of the protocol, the testing was conducted over two separate testing sessions. Each participant performed the identical testing procedure on two separate occasions that was separated by five to nine days (ideally 7 days). Participants were also tested at similar times of the day in order to minimise the influence of diurnal variation influences.

4.4 Data processing
4.4.1 EMG processing

a. *Removal of movement artefacts*. The data were exported as a C3D to MATLAB for processing with custom written software in MATLAB. There were three steps to process the raw EMG data. The first step used a high-pass filter (20Hz for walking and 30Hz for running trials) to remove movement artefacts and noise, as the typical frequency range of cable motion artefacts is between 1 and 50 Hz (Clancy, Morin, &

Merletti, 2002). The second step was rectification and envelope detection, which made all signals positives. The final step was a low- pass filter (6Hz), which was used by (Hubley-Kozey, Deluzio, Landry, McNutt, & Stanish, 2006; Hubley-Kozey, Hatfield, Wilson, & Dunbar, 2010; Winter & Yack, 1987) to create a linear envelope, as recommended for EMG processing for dynamic tasks (Hermens et al., 1999). Using a filter of 6Hz maintained at least 95% of signal power (Shiavi, Frigo, & Pedotti, 1998). Following EMG processing, the data were exported to a Microsoft Excel 2016 spreadsheet to obtain a final result.

b. *Time normalisation*. For each gait cycle, EMG data were normalised to 100% of the stance phase of the gait cycle before export within MATLAB. The gait events foot strike and toe off were defined using the force data, as explained in section 4.3.5.4. With these data it was possible to define the precise sample corresponding to 0% and 100% of the stance phase of the gait cycle. However, muscles are active before, and after, stance phase and therefore a window of 50% of the stance phase was included before and after the 0-100% window. The samples corresponding to -50% and 150% were defined by subtracting/adding 50% of the stance phase time to the 0% and the 100% time points. Interpolation of the EMG signals was then performed from -50% to 150% using the Matlab function interp1.

c. *Data averaging*. The data from ten gait cycles that had consistent kinematic patterns and were within the normal range (Pinzone, Schwartz, Thomason, & Baker, 2014) were averaged to produce an individual ensemble average EMG profile.

4.4.1.1 Dynamic EMG data

Time normalisation: for each gait cycle, EMG and kinematic data were normalised to 100% of the stance phase of the gait cycle.

4.4.1.2 Calculation of reference amplitudes (normalisation techniques)

MVIC

As mentioned before in the testing protocol section (4.3.7), the participants were asked to perform three maximum contractions each of the tested muscles for three seconds. They were instructed to push as hard as they could in each position against the manual resistance with similar verbal encouragement throughout the whole testing process. A minimum of a minute was given as rest period between each muscle contraction to eliminate the effect of fatigue. The recorded EMG data were processed as follows:

a. Each recorded EMG was processed typically, as mentioned in 4.4.1 (a-b).

b. The average RMS EMG signal was calculated separately from the middle one second of each of the three MVICs.

c. The largest of the three values was chosen as the MVIC normalisation factor for each muscle for each individual during each testing session.

d. The MVIC was then used to normalise the ensemble average. This process resulted in creating a normalised EMG ensemble average to MVIC for each muscle (Figure 4-10b-c).

e. This process was repeated for all tested muscles for each testing day.

Mean and peak of the dynamic trials (MDT and PDT)

a. Each recorded EMG signal from the dynamic trials was processed in a similar way to what was described in 4.5.1.

b. Based on the ensemble average obtained from the dynamic tasks for each muscle for each testing session, the mean and peak of the dynamic trial values were calculated.

c. The mean and peak amplitudes of the dynamic trials were then used to normalise the ensemble average. This process resulted in creating a normalised EMG ensemble average to mean dynamic and peak dynamic for each muscle (see Figure 4- 10d-e).

d. This process was repeated for all tested muscles for each testing day.
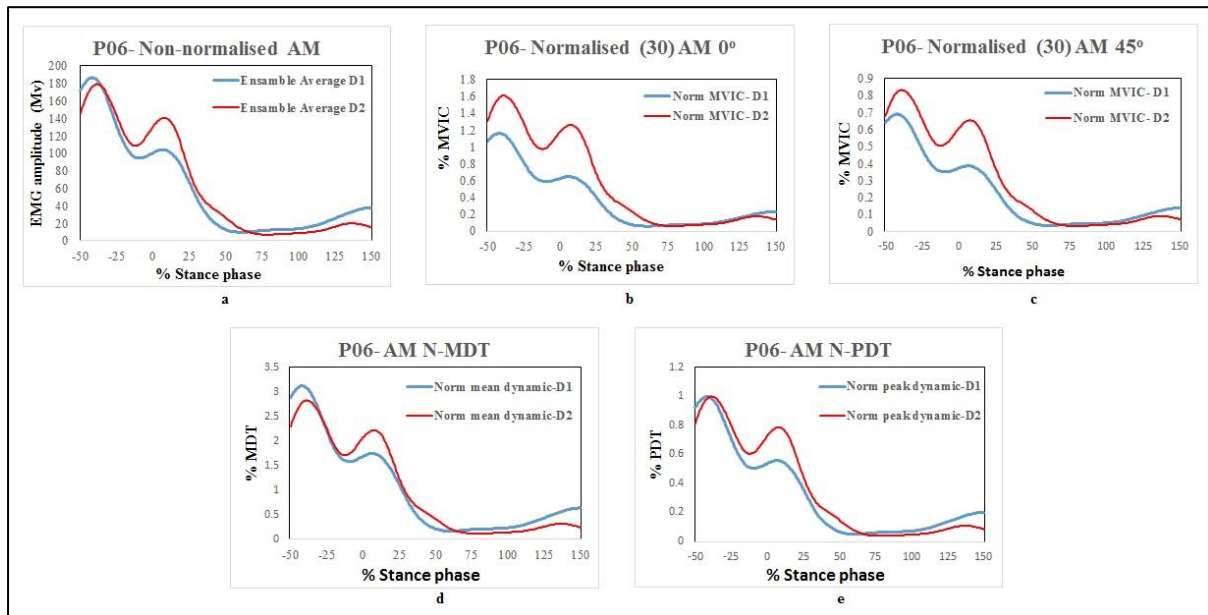
Figure 4- 10: An example of the processing of EMG data for the adductor magnus muscle (AM) during the stance phase of walking. a. The non-normalised EMG data obtained from AM during the stance of walking on two different occasions, day 1 (blue line), and day 2(red line); b. Normalised EMG data to MVIC value obtained at neutral hip position; c. Normalised EMG data to MVIC value obtained at 45º hip flexion angle; d. Normalised EMG data to mean value of the dynamic trials for each testing day; e. normalised EMG data to peak value of the dynamic trials for each testing day.

4.4.2 Three-dimensional processing

4.4.2.1 Modelling

Kinematics and kinetics data were processed using Qualisys Track Manager (QTM) software (Qualisys AB, Partille, Sweden) for marker labelling and filling any trajectory gaps with maximum consecutive ten frames. After that, the data files were exported to Visual 3D (V3D) software (Version 6, C-motion Incorporation, Germantown, USA) as coordinate 3D (C3D) files to manage, analyse, and report the related data. Visual3D motion was used to calculate

joint kinematic and kinetic data. Walking and running motion data were filtered using a Butterworth 4th order bi-directional low-pass filter with cut-off frequencies of 6Hz and 12Hz, respectively (Winter, 2009; Yu, Gabriel, Noble, & An, 1999). All lower-extremity segments were modelled as conical frusta, with inertial parameters estimated from anthropometric data (Dempster, Gabel, & Felts, 1959).

There are different ways in which joints and segments can be defined. The most commonly used systems are the global coordinate system (GCS), segment coordinate systems (SCS), and joint coordinate systems (JCS). The GCS is where the segment angles are calculated from the x, y, and z axes of the laboratory. The SCS uses the proximal and distal endpoints of the segment to determine an orientation of the x, y, and z axes of the joint. The JCS is where the axis of the two body segments (shank and thigh for example) is used to create a third floating axis. Joint angles are calculated in Visual 3D using the Cardan/Euler representation which finds the orientation of the distal segment with respect to the reference proximal segment using the x, y, z sequence of rotations (Cole, Nigg, Ronsky, & Yeadon, 1993). The axes are defined as X that describes flexion-extension, Y describes abduction-adduction/varus-valgus, and Z describes internal-external rotation. This sequence of rotations and coordinate system definitions has been shown to be equivalent to the 'floating axis' system (Grood & Suntay, 1983). Using a rotation sequence of order x, y, z for light-handed coordinate systems, the first, second and third angular displacements resulted in flexion-extension, abduction-adduction and internal-external rotation at the hips and knees, and dorsiflexion-plantarflexion, eversion-inversion and internal-external rotation at the ankles, respectively. The zero positions of all joint angles during the dynamic trials were derived when the subjects stood comfortably in the static test position. The investigation enabled the calculation of kinematic and kinetic data in all three of the anatomic planes within the reference system for both lower limbs. The subject's body mass (in kilograms) was entered

into the software for use in kinetic calculations, i.e. was used to normalise the joint moment data.

The calibration anatomical systems technique (CAST) was used to define the six degrees of freedom movement of each segment during the dynamic tasks (Cappozzo et al., 1996). A static trial, during which the participant stood on the force plates with all markers in view of the cameras, was done with all the anatomical and tracking markers and the Qualisys software prior to extraction for post-processing software. The positions of these anatomical markers offered reference points to identify bone movement through only the tracking markers set during the movement trials.

As can be seen in Figure 4- 11 and Table 4- 2, the model used had seven rigid segments attached to the joint. Each segment is considered to have six variables that describe its position (three variables describe the position of the origin, and three variables describe the rotation) in 3D space. Specifically, three variables describe the segment translation along three perpendicular axes (vertical, medial-lateral, and anterior-posterior), and three variables describe the rotation about each axis of the segment (sagittal, frontal, and transverse). Each segment of the pelvis, thigh, shank, and foot was modelled to determine the proximal and distal joint. The local coordinate systems (x, y, z) were defined by specifying a medial and lateral location at the proximal and distal ends of the segment with the segment endpoints being the mid-point between these two locations (e.g., mid-point between lateral and medial femoral condyles).

Figure 4- 11: A. QTM static model; B. V3D^{TM} standing bone model. Red = x-axis, Green = y-axis, and Blue = z-axis. Right iliac crest (RIliac), right anterior superior iliac spin (RASIS), right greater trochanter (RGTroc), right lateral femoral condyle (R Lat Condyle), right medial femoral condyle (R Med Condyle), right lateral malleolus (R Lat Malleolus), right medial malleolus (R Med Malleolus), right fifth metatarsal head (R 5MT), right first metatarsal head (R 1MT).

Markers for the segment motion definition

Pelvis segment:

For the pelvis segment the Cartesian Optoelectronic Dynamic Anthropometer (CODA) model was used. The pelvis segment and the hip joint centre were defined using the anatomical locations of the ASIS and the PSIS on both sides. The x-y plane of the segment coordinate system was defined as the plane passing through the right and left ASIS markers and the mid-point of the right and left PSIS markers and the z-axis is perpendicular to the (x-y) plane. The hip joint centre was defined by the distance between the right and left ASIS markers. This technique was firstly introduced by Bell, Brand, and Pedersen (1989) and confirmed by

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Reize, Muller, Motzny, and Wulker (2006). The location of the hip joint centre was predicted using external landmarks (i.e. right and left ASIS). Consequently, the hip joint centre in adults (expressed as percentage of the distance between the ASISs) was determined as 14% medially, 30% distally, and 22% posteriorly to the anterior superior iliac spine. The z axis was then defined as the line passing through the ASISs with its positive direction from left to right. The x axis was then defined by the line from the midpoint of the PSISs to the midpoint of the ASISs with its positive direction forwards. Finally, the y axis was defined to be perpendicular to the xz plane with its positive direction is proximal (Wu et al., 2002).

Thigh, leg and foot segments:

A thigh cluster and a leg cluster (Figure 4- 12), each includes four markers, were placed over the thigh and leg using double sided adhesive tape and secured with additional creep bandage on right and left side of the lower extremities. The thigh origin was coincident with the right (or left) hip centre of rotation, coincident with that of the pelvic coordinate system (O) in the neutral configuration (Figure 4- 5). The y axis joins the origin and medial and lateral femoral epicondyles and its positive direction is proximal. The z axis was defined as the line perpendicular to the y-axis, lying in the plane defined by the origin and the two femoral epicondyles, pointing to the right. The x axis is perpendicular to the yz axes with its positive direction forwards (Wu et al., 2002).

The shank (leg) origin was located at the midpoint of the line joining the lower ends of the malleoli (Figure 4- 13). The mid points of the malleoli and the epicondyles were used to define the proximal and distal end of the shank segment respectively. The z axis was defined as the line connecting the two malleoli and pointing to the right. The x axis was defined as the line perpendicular to the torsional plane of the tibia/fibula and pointing anteriorly. The y axis is perpendicular to the xz axes with its positive direction proximal (Wu et al., 2002).

Figure 4- 12: Cluster plate for thigh and leg segments.
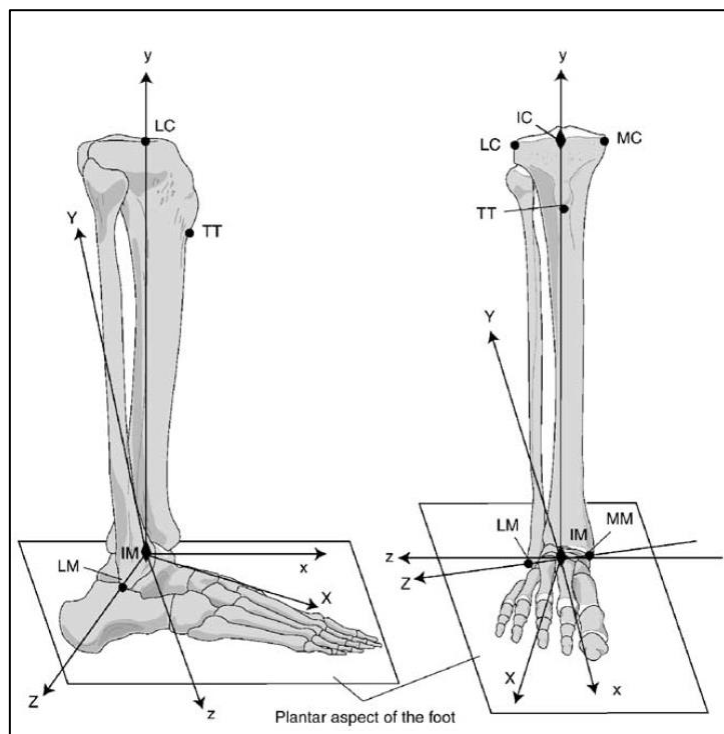
Figure 4- 13: Illustration of the tibia/fibula coordinate system (XYZ) and the calcaneus coordinate system (xyz) with the ankle joint complex in the neutral position. Where, MM: Tip of the medial malleolus; LM: Tip of the lateral malleolus; MC: The most medial point on the border of the medial tibial condyle; LC: The most lateral point on the border of the lateral tibial condyle; TT: Tibial tuberosity. IM: The inter-malleolar point located midway between MM and LM; and IC: The inter-condylar point located midway between the MC and LC. (Adopted from (Wu et al., 2002).

139

The ankle joint complex is composed of the talocrural and the subtalar joints. Since the ankle joint and the toe targets were not parallel to the floor, an offset was caused in the ankle angle which resulted an increased plantar flexion. To remove this offset, a left and right virtual foot were built to create a clinically relevant ankle joint angle. Therefore, the neutral ankle angle was defined as a flat foot with a vertical shank segment, regardless of the actual foot posture during the static trial. The proximal markers for the foot segment were the medial and lateral malleolus and the distal landmarks were the 1st and 5th toe. The four markers were used to build the neutral virtual foot and to calculate the ankle kinematics.

Markers were placed over the standard shoe (Figure 4- 7) with glue to ensure no motion of the markers during the dynamic trials. The markers placement allowed the subsequent analysis of the motion of the foot during gait. In order to minimise any movement of the foot inside the shoe, a firm fitting of the shoes for each participant was made. Therefore, the segment's origin was located at the midpoint of the malleoli (Figure 4- 13). The z axis was defined as the line connecting medial and lateral malleoli and pointing to the right. The x axis was perpendicular to the torsional plane of the tibia/fibula and pointing anteriorly. The y axis was the common line perpendicular to xz axes (Wu et al., 2002).

Table 4- 2: V3D model, with seven main segments.

Segment name	Segment type	Segment markers
Pelvis	Coda	• Anterior: right and left ASIS • Posterior: right and left PSIS
Right and left thigh	V3D	• Joint centre: hip joint • Distal: medial and lateral knee epicondyles
Right and left shank	V3D	• Proximal: medial and lateral knee epicondyles • Distal: medial and lateral ankle malleolus
Right and left foot	V3D	• Proximal: medial and lateral ankle malleolus • Distal: first- and fifth-foot metatarsals

Table 4- 3: The pipeline used to analysis the V3D data with eight automatic sequential processing commands.

	Command	Command preview
1	File open	To open the motion (walking or running) files
2	Create hybrid model	To open the standing (static) file
3	Apply model template	To open the study model file (i.e. model that was created for the purpose of this study (see above Table 4- 2).
4	Assign model file	To select the participant's trials and apply it into the study model
5	Set the subject height	To enter subject height and measured unit
6	Set the subject height	To enter subject weight and measured unit
7	Interpolate	To fill any trajectory gap (i.e. missing data of a signal) with maximum ten frames of polynomial interpolation type
8	Low-pass filter	To digitally filter any low frequency noise with a cut-off frequency of 12Hz for kinematics data
9	Low-pass filter	To digitally filter any high frequency noise with a cut-off frequency of 25Hz for kinetics data
10	Automatic gait events	To show and edit events' dialog for each motion trial while using automatic labelling at heel strike (foot on) and toe off (foot off)
11	Report template	To display the descriptive information for each participant (i.e. name, ID, height, weight, sex, date of birth, test date, diagnosis, and test conditions) and to show the graphs of the desired joint motion angles.

4.5 Statistical analysis

All statistical analysis was carried out using SPSS for Windows version 23.0 (SPSS Inc.,

Chicago, IL) and Excel (Microsoft Office Excel, 2016). The coefficient of multiple

correlation (CMC) was used to assess the EMG consistency between days. This test sets the

degree of waveform similarity between different testing sessions. CMC has been used

frequently in the previous studies to assess the reliability of EMG measurement (Growney et al., 1997; Kadaba et al., 1989). CMC affords a value between 0 and 1 regardless the unit of measurement. The closer the value to 1, the better the agreement between testing sessions (Collins, Ghoussayni, Ewins, & Kent, 2009; Growney et al., 1997). Values less than 0.5 are indicative of poor reliability, values between 0.5 and 0.75 indicate moderate reliability, values between 0.75 and 0.9 indicate good reliability, and values greater than 0.90 indicate excellent reliability (Portney, 2009).

Although the CMC seems to be easy to interpret, it alone cannot provide a comprehensive overview of reliability and should be complemented by confidence intervals (CI). Additionally, the CMC does not provide any indication of the amount of agreement between measurements, in the unit of the actual measurement. Therefore, standard error of measurement (SEM) was used in conjunction with the CMC and a CI of 95%.

Calculation of SEM was done using the formula: $SEM = \sqrt{\Sigma\ \text{deviation}^2\ /\ \text{degree of freedom}}$ (Bland & Altman, 1996). The SEM values are presented in the same units of the measurement of the measured variable (degrees for joint angles, percent of normalisation method for EMG amplitudes) (Blankevoort, van Heuvelen, & Scherder, 2013; Bruton, Conway, & Holgate, 2000). One-way repeated measures analysis of variance (ANOVA) was used to compare CMC values among the three normalisation methods (MVIC, MDT, and PDT). In order to calculate the standard error of measure (SEM), the peak values of the kinematic variables (including pelvis, hip and knee angles) were measured at the period of interest (between 0-20% of the gait cycle) where CPD is most likely to occur. This period represents 20% of the gait cycle (i.e. equal to 33% and 67% of the stance phase of walking and running; respectively). An ANOVA with repeated measures is usually used to compare three or more group means where the participants are the same in each group. This usually

occurs when participants are subjected to more than one testing condition and the response to each of these conditions is investigated. In the current study, each muscle's activity for the same participant was normalised by three different techniques (MVIC, MDT, and PDT). Therefore, the best choice to compare between the different normalisation methods is the One-way repeated measures ANOVA. The level of significance was set at 0.05. Additional analysis was performed to understand possible changes in MVIC between different test days. For this testing, the consistency of the absolute value (raw signal) of the MVIC for each muscle was used. Note that because different filtering frequencies were used to process the walking and running data, this analysis was carried out separately for walking and running using the data raw data processed in the two different ways. This calculation was performed using an ICC and SEM. The guide stated above was used to interpret these values. Specifically, an ICC less than 0.5 were taken to be indicative of poor reliability, values between 0.5 and 0.75 taken to indicate moderate reliability, values between 0.75 and 0.9 to indicate good reliability and values greater than 0.90 to indicate excellent reliability (Portney, 2009).

4.6 Results
4.6.1 Repeatability of EMG measurements

A cohort of ten healthy male runners who had no history of lower-limb injury or surgery participated in this study. The mean (SD) age of the subjects was 30 (7.1) years, mean height 1.74 (0.1) m, mean weight 70.1 (7.5) kg, and the mean body mass index (BMI) 23.1 (1.4) kg/m^2.

For the walking task, EMG activity in all tested muscles (GMax, GMed, VMO, VLO, TA, MGastro, and LGastro) exhibited excellent repeatability apart from MHam, LHam, AM, AL, and Gr. The different methods of normalisation exhibit different levels of repeatability. The EMG data for GMax, GMed, VMO, VLO, TA, MGastro, and LGastro exhibited excellent

repeatability when normalised to MVIC (Table 4- 4), normalised to MDT (Table 4- 5), and to PDT (Table 4- 6). Considering the different methods of normalisations, the other tested muscles (MHam, LHam, AM, AL, and Gr) produced different values of good repeatability (0.75-0.88) with slight lower CMC values when normalised to MVIC.

For the running task, EMG activity of GMax, GMed, VMO, VL, AM, MGastro, and LGastro exhibited excellent repeatability apart from MHam, LHam, AL, Gr, and TA. Similar to walking, the different methods of normalisation exhibit different levels of repeatability. The EMG data for GMax, GMed, VMO, VLO, AM, MGastro, and LGastro exhibited excellent repeatability when normalised to MVIC (Table 4- 7), normalised to MDT (Table 4- 8) and to PDT (Table 4- 9). However, the MHam and TA exhibited good repeatability when normalised to MVIC, they produced excellent repeatability when normalised to MDT and PDT. The LHam and AL exhibited good repeatability regardless the method of normalisation, however; higher repeatability values produced when normalised to MDT and PDT. Additionally, the Gr activities exhibited moderate repeatability regardless the method of normalisation, however; higher repeatability values produced when normalised to MDT and PDT.

Table 4- 4: Mean and standard deviation of the coefficient of multiple correlation (CMC) and the standard error of measurement (SEM) of EMG data obtained during walking. The EMG data were normalised to MVIC.

Muscle	CMC		SEM				Average Peak
	Average	SD	Value	CI	Upper bound	Lower bound	
GMax	0.93	0.04	0.02	0.002	0.021	0.018	0.22
GMed	0.93	0.03	0.04	0.006	0.174	0.035	0.44
VMO	0.90	0.05	0.03	0.004	0.033	0.025	0.25
VLO	0.93	0.05	0.03	0.004	0.033	0.026	0.36
AL	0.75	0.09	0.02	0.001	0.022	0.019	0.09
Gr	0.76	0.08	0.03	0.003	0.036	0.030	0.20
AM	0.87	0.11	0.20	0.036	0.241	0.169	0.64
TA	0.92	0.03	0.04	0.002	0.041	0.037	0.34
MHam	0.80	0.09	0.09	0.016	0.104	0.072	0.31
LHam	0.76	0.15	0.03	0.005	0.035	0.025	0.23
MGastro	0.90	0.03	0.05	0.009	0.061	0.044	0.57
LGastro	0.94	0.03	0.02	0.003	0.022	0.016	0.38

Table 4- 5: Mean and standard deviation of the coefficient of multiple correlation (CMC) and the standard error of measurement (SEM) of EMG data obtained during walking. The EMG data were normalised to the mean of the dynamic trials (MDT).

Muscle	CMC		SEM				Average Peak
	AVERAGE	SD	Value	CI	Upper bound	Lower bound	
GMax	0.94	0.04	0.25	0.03	0.27	0.22	5.38
GMed	0.95	0.02	0.16	0.02	0.17	0.14	4.69
VMO	0.95	0.03	0.21	0.02	0.23	0.19	4.39
VLO	0.97	0.01	0.19	0.02	0.21	0.17	4.52
AL	0.78	0.03	0.33	0.02	0.35	0.31	2.93
Gr	0.83	0.07	0.27	0.01	0.29	0.26	3.04
AM	0.89	0.09	0.31	0.03	0.34	0.28	3.84
TA	0.94	0.03	0.14	0.01	0.15	0.14	3.41
MHam	0.88	0.06	0.29	0.03	0.31	0.26	3.73
LHam	0.78	0.14	0.27	0.03	0.30	0.24	4.56
MGastro	0.93	0.01	0.19	0.02	0.21	0.17	5.31
LGastro	0.96	0.01	0.22	0.02	0.25	0.20	5.71

Table 4- 6: Mean and standard deviation of the coefficient of multiple correlation (CMC) and standard error of measurement (SEM) of EMG data obtained during walking. The EMG data were normalised to the peak of the dynamic trials (PDT).

Muscle	CMC		SEM			
	AVERAGE	SD	Value	CI	Upper bound	Lower bound
GMax	0.96	0.01	0.05	0.004	0.06	0.05
GMed	0.95	0.02	0.04	0.003	0.04	0.03
VMO	0.95	0.04	0.06	0.003	0.07	0.06
VLO	0.93	0.06	0.05	0.004	0.05	0.04
AL	0.83	0.09	0.13	0.01	0.14	0.12
Gr	0.83	0.09	0.19	0.02	0.21	0.17
AM	0.89	0.09	0.10	0.004	0.10	0.09
TA	0.94	0.02	0.05	0.003	0.05	0.05
MHam	0.88	0.06	0.09	0.01	0.10	0.08
LHam	0.77	0.15	0.09	0.01	0.09	0.08
MGastro	0.97	0.02	0.04	0.004	0.04	0.04
LGastro	0.96	0.01	0.04	0.004	0.05	0.04

Table 4- 7: Mean and standard deviation of the coefficient of multiple correlation (CMC) and the standard error of measurement (SEM) of EMG data obtained during running. The EMG data were normalised to MVIC.

Muscle	CMC		SEM				Average Peak
	Average	SD	Value	CI	Upper bound	Lower bound	
GMax	0.94	0.02	0.14	0.02	0.16	0.11	0.80
GMed	0.91	0.06	0.16	0.02	0.18	0.04	0.92
VMO	0.91	0.07	0.19	0.03	0.17	0.17	1.26
VLO	0.94	0.03	0.15	0.02	0.17	0.13	1.38
AL	0.83	0.11	0.10	0.01	0.11	0.10	0.36
Gr	0.70	0.11	0.21	0.02	0.23	0.18	0.72
AM	0.91	0.04	0.11	0.01	0.12	0.09	0.95
TA	0.87	0.08	0.21	0.01	0.08	0.07	0.66
MHam	0.87	0.07	0.30	0.03	0.33	0.27	0.71
LHam	0.77	0.17	0.17	0.02	0.19	0.16	0.59
MGastro	0.95	0.03	0.22	0.03	0.24	0.19	1.27
LGastro	0.96	0.02	0.07	0.01	0.08	0.06	0.93

Table 4- 8: Mean and standard deviation of the coefficient of multiple correlation (CMC) and the standard error of measurement (SEM) of EMG data obtained during running. The EMG data were normalised to the mean of the dynamic trials (MDT).

Muscle	CMC		SEM				Average Peak
	AVERAGE	SD	Value	CI	Upper bound	Lower bound	
GMax	0.94	0.02	0.16	0.02	0.18	0.15	3.55
GMed	0.96	0.02	0.16	0.02	0.18	0.15	3.56
VMO	0.97	0.02	0.14	0.02	0.21	0.17	3.68
VLO	0.97	0.01	0.13	0.02	0.20	0.17	3.49
AL	0.88	0.06	0.32	0.02	0.34	0.31	2.21
Gr	0.74	0.12	0.34	0.03	0.36	0.31	2.44
AM	0.93	0.02	0.33	0.03	0.36	0.31	2.84
TA	0.92	0.05	0.20	0.01	0.16	0.15	2.35
MHam	0.92	0.05	0.25	0.02	0.31	0.27	3.19
LHam	0.83	0.12	0.20	0.01	0.25	0.22	2.56
MGastro	0.97	0.01	0.13	0.02	0.21	0.17	3.42
LGastro	0.96	0.02	0.17	0.01	0.18	0.15	3.46

Table 4- 9: Mean and standard deviation of the coefficient of multiple correlation (CMC) and the standard error of measurement (SEM) of EMG data obtained during running. The EMG data were normalised to the peak of the dynamic trials (PDT).

Muscle	CMC		SEM			
	AVERAGE	SD	Value	CI	Upper bound	Lower bound
GMax	0.95	0.02	0.34	0.04	0.39	0.30
GMed	0.96	0.02	0.04	0.004	0.05	0.04
VMO	0.97	0.01	0.04	0.004	0.04	0.04
VLO	0.97	0.01	0.04	0.003	0.04	0.04
AL	0.85	0.11	0.15	0.01	0.16	0.14
Gr	0.73	0.12	0.17	0.01	0.17	0.16
AM	0.93	0.03	0.11	0.01	0.11	0.10
TA	0.92	0.05	0.07	0.003	0.07	0.06
MHam	0.92	0.05	0.09	0.01	0.09	0.08
LHam	0.82	0.12	0.10	0.01	0.10	0.09
MGastro	0.98	0.01	0.04	0.005	0.05	0.04
LGastro	0.97	0.01	0.06	0.003	0.06	0.05

4.6.2 Effects of normalisation techniques on between-session reliability
4.6.2.1 Effects of normalisation techniques on between-session reliability during walking

During walking, only the VMO, MHam, and MGastro muscles were affected by the different normalisation methods. Specifically, peak and mean normalisations reduced SEM and increased CMC compared to MVIC normalisation. The mean normalisation CMC value of VMO was significantly higher than the value from MVIC; however, the mean and peak normalisations of MHam and MGastro were significantly higher than those from MVIC. Similar behaviour was observed for remaining tested muscles (AL, GR, GMax, GMed, VLO, TA, LHam and LGastro) which had a non-significant increase in PDT and/or MDT normalisations values. These results are shown in Table 4- 10.

Table 4- 10: Comparison of mean (SD) of CMC values using repeated measures ANOVA for EMG amplitude normalised to different normalisation methods for lower limb muscles during walking.

Muscle	Normalisation methods during walking		P value
GMax	MVC: 0.93 (0.04)	MDT	0.804
		PDT	0.200
	MDT: 0.94 (0.04)	PDT	0.504
	PDT: 0.96 (0.01)	MDT	0.504
GMed	MVC: 0.93 (0.03)	MDT	0.188
		PDT	0.188
	MDT: 0.95 (0.02)	PDT	1.000
	PDT: 0.95 (0.02)	MDT	1.000
VMO	MVC: 0.90 (0.05)	MDT	**0.049***
		PDT	0.078
	MDT: 0.95 (0.03)	PDT	0.971
	PDT: 0.95 (0.04)	MDT	0.971
VLO	MVC: 0.93 (0.05)	MDT	0.326
		PDT	0.990
	MDT: 0.97 (0.01)	PDT	0.372
	PDT: 0.93 (0.06)	MDT	0.372
AL	MVC: 0.75 (0.09)	MDT	0.829
		PDT	0.243
	MDT: 0.78 (0.03)	PDT	0.511
	PDT: 0.83 (0.09)	MDT	0.511
Gr	MVC: 0.76 (0.08)	MDT	0.214
		PDT	0.343
	MDT: 0.83 (0.07)	PDT	0.952
	PDT: 0.82 (0.09)	MDT	0.952
AM	MVC: 0.87 (0.11)	MDT	0.989
		PDT	0.967
	MDT: 0.89 (0.09)	PDT	0.994
	PDT: 0.89 (0.09)	MDT	0.994
TA	MVC: 0.92 (0.03)	MDT	0.371
		PDT	0.396
	MDT: 0.94 (0.03)	PDT	1.000
	PDT: 0.94 (0.02)	MDT	1.000
MHam	MVC: 0.80 (0.09)	MDT	**0.034***
		PDT	**0.037***
	MDT: 0.88 (0.06)	PDT	0.999
	PDT: 0.88 (0.06)	MDT	0.999
LHam	MVC: 0.76 (0.15)	MDT	0.963
		PDT	0.984
	MDT: 0.78 (0.14)	PDT	0.995
	PDT: 0.77 (0.15)	MDT	0.995
MGastro	MVC: 0.90 (0.03)	MDT	**0.024***
		PDT	**0.022***
	MDT: 0.93 (0.01)	PDT	0.999
	PDT: 0.97 (0.02)	MDT	0.999
LGastro	MVC: 0.94 (0.03)	MDT	0.107
		PDT	0.136
	MDT: 0.96 (0.01)	PDT	0.992
	PDT: 0.96 (0.01)	MDT	0.992

*P< 0.05

4.6.2.2 Effects of normalisation techniques on between-session reliability during running

In the same manner, the peak and mean normalisations reduced SEM and increased CMC during running compared to MVIC normalisation for selected muscles. The EMG profiles show that the GMed, VMO, VLO, MHam, and MGastro muscles are affected by the different normalisation methods. The MVIC normalisation CMC values of GMed, VMO, and VLO were significantly lower than the peak and mean values; however, the MVIC normalisations of MHam and MGastro were significantly lower than those from the peak. Similar behaviour was observed for the remaining tested muscles (AL, GR, GMax, GMed, TA, LHam, and LGastro), which had a non-significant increase in PDT and/or MDT normalisations values. These results are shown in Table 4- 11.

Table 4- 11: Comparison of mean (SD) of CMC values using repeated measures ANOVA for EMG amplitude normalised to different normalisation methods for lower limb muscles during running.

Muscle	Normalisation methods during running		P value
GMax	MVC: 0.94 (0.02)	MDT	0.803
		PDT	0.431
	MDT: 0.94 (0.02)	PDT	0.837
	PDT: 0.95 (0.02)	MDT	0.837
GMed	MVC: 0.91 (0.06)	MDT	**0.015***
		PDT	**0.015***
	MDT: 0.96 (0.02)	PDT	1.000
	PDT: 0.96 (0.02)	MDT	1.000
VMO	MVC: 0.91 (0.07)	MDT	**0.008***
		PDT	**0.009***
	MDT: 0.97 (0.02)	PDT	0.998
	PDT: 0.97 (0.01)	MDT	0.998
VLO	MVC: 0.94 (0.03)	MDT	**0.021***
		PDT	**0.019***
	MDT: 0.97 (0.01)	PDT	0.994
	PDT: 0.97 (0.01)	MDT	0.994
AL	MVC: 0.83 (0.11)	MDT	0.614
		PDT	0.906
	MDT: 0.88 (0.06)	PDT	0.843
	PDT: 0.85 (0.11)	MDT	0.843
Gr	MVC: 0.70 (0.11)	MDT	0.652
		PDT	0.793
	MDT: 0.74 (0.12)	PDT	0.975
	PDT: 0.73 (0.12)	MDT	0.975
AM	MVC: 0.91 (0.04)	MDT	0.232
		PDT	0.395
	MDT: 0.93 (0.02)	PDT	0.936
	PDT: 0.93 (0.03)	MDT	0.936
TA	MVC: 0.87 (0.08)	MDT	0.166
		PDT	0.166
	MDT: 0.92 (0.05)	PDT	1.000
	PDT: 0.92 (0.05)	MDT	1.000
MHam	MVC: 0.87 (0.07)	MDT	0.056
		PDT	**0.036***
	MDT: 0.92 (0.05)	PDT	0.979
	PDT: 0.92 (0.05)	MDT	0.979
LHam	MVC: 0.77 (0.17)	MDT	0.611
		PDT	0.718
	MDT: 0.83 (0.12)	PDT	0.983
	PDT: 0.82 (0.12)	MDT	0.983
MGastro	MVC: 0.95 (0.03)	MDT	0.055
		PDT	**0.020***
	MDT: 0.97 (0.01)	PDT	0.915
	PDT: 0.98 (0.01)	MDT	0.915
LGastro	MVC: 0.96 (0.02)	MDT	0.603
		PDT	0.133
	MDT: 0.96 (0.02)	PDT	0.532
	PDT: 0.97 (0.01)	MDT	0.532

*P< 0.05

4.6.2.3 Reliability of MVIC test

The ICC values of the MVIC were computed in order to understand the variability of MVIC between the two testing days calculated using the filtering methods used for walking (Table 4- 12). The muscles VMO, VLO and TA showed moderate reliability and the GMed, Gr, MHam, MGastro and LGastro showed good reliability. Additionally, the GMax, AM and LHam show excellent reliability. However, for one muscle the AL, the ICC values were low showing poor reliability and high variability between the testing days.

The corresponding ICC for the MVIC data which was used to normalise the EMG amplitudes for running are displayed in Table 4- 13. These data show that the MVIC values for the AL and VLO show poor reliability. In addition, the TA shows moderate reliability but the GMed, VMO, Gr, MGastro and LGastro show good reliability. As with walking, the GMax, AM and MHam show excellent reliability.

Table 4- 12:Between-days ICC, SEM, mean, standard deviation and the peak signal during MVIC data used to normalise the walking task

Muscle	ICC	SEM	Mean	SD	Peak signals
GMax	0.93	80.87	305.95	20.07	728.92
GMed	0.78	91.05	265.86	44.91	605.09
VMO	0.61	117.18	292.52	7.63	635.91
VLO	0.74	65.01	198.60	35.78	439.65
AL	0.35	203.24	450.87	26.94	965.59
Gr	0.77	69.80	201.31	11.25	487.55
AM	0.99	11.82	103.48	2.34	263.51
TA	0.62	147.23	454.14	1.71	943.77
MHam	0.85	96.85	362.82	29.54	728.75
LHam	0.94	47.02	283.65	8.99	511.69
MGastro	0.81	84.30	427.56	22.98	537.15
LGastro	0.89	82.20	375.49	29.09	756.73

Table 4- 13: Between-days ICC, SEM, mean, standard deviation and the peak signal for the MVIC data used to normalise the running task

Muscle	ICC	SEM	Mean	SD	Peak signals
GMax	0.95	71.18	300.89	11.59	666.35
GMed	0.79	79.28	254.56	28.50	472.27
VMO	0.78	85.39	272.712	21.08	635.91
VLO	0.48	78.52	179.64	33.03	430.13
AL	0.38	133.42	438.53	2.88	683.10
Gr	0.90	45.06	198.08	1.83	384.57
AM	0.98	12.03	99.15	3.33	221.86
TA	0.54	174.47	474.85	31.07	943.77
MHam	0.93	73.57	376.13	12.88	728.75
LHam	0.90	53.40	281.10	3.98	520.87
MGastro	0.83	84.83	365.73	12.66	681.91
LGastro	0.86	94.51	360.54	11.58	849.74

4.6.3 Reliability of kinematics measurements

The CMC and SEM values for walking and running are shown in Table 4- 14 and Table 4-15, respectively. For sagittal plane movements, the repeatability of joint angle motion at the hip and knee were excellent both in walking and running. Similarly, the frontal plane movements at pelvis and hip displayed excellent between days repeatability for both tasks. The CMCs for the joint angle motion in the frontal plane (0.90-0.96) were slightly lower than those for sagittal plane motion (0.97-0.99) but still excellent. The CMC exceeded 0.96 in walking for all movements apart from pelvic tilt (0.94) which still excellent. For running data, the CMC value in the sagittal plane was the highest value (0.97-0.99). In contrast, the CMCs for hip and pelvis in the frontal plane were relatively low (0.90-0.92) but still excellent.

Table 4- 14: Mean and standard deviation of the coefficient of multiple correlation (CMC) and the standard error of measurement (SEM) during walking.

Joint	CMC		SEM	Peak angle (IC to 33% of stance cycle= 20% of the gait cycle)
	Average	SD		
R-Pelvis-Frontal	0.94	0.046	1.080	-8.35
L-Pelvis-Frontal	0.94	0.046	0.713	-10.27
R-Hip- Sagittal	0.99	0.008	2.077	36.34
L-Hip- Sagittal	0.99	0.005	2.262	37.77
R-Hip-Frontal	0.96	0.024	1.988	12.19
L-Hip-Frontal	0.96	0.023	1.180	14.08
R-Knee-Sagittal	0.99	0.007	2.002	18.99
L-Knee- Sagittal	0.99	0.003	1.084	18.39

Table 4- 15: Mean and standard deviation of the coefficient of multiple correlation (CMC) and the standard error of measurement (SEM) during running.

Joint	CMC		SEM	Peak angle (IC to 67% of stance cycle= 20% of the gait cycle)
	Average	SD		
R-Pelvis-Frontal	0.90	0.06	0.889	-8.32
L-Pelvis-Frontal	0.92	0.05	0.673	-7.88
R-Hip- Sagittal	0.98	0.02	2.302	40.43
L-Hip- Sagittal	0.99	0.01	2.470	44.99
R-Hip-Frontal	0.92	0.06	1.932	12.66
L-Hip-Frontal	0.92	0.05	1.792	15.77
R-Knee-Sagittal	0.97	0.04	1.506	45.27
L-Knee- Sagittal	0.99	0.01	2.179	49.88

4.7 Discussion
4.7.1 Repeatability of EMG

The results of this study demonstrate that most variables of interest exhibited high repeatability in both walking and running. During the walking task, EMG activity exhibited excellent repeatability in the majority of the tested muscles (GMax, GMed, VMO, VL, AM, TA, MGastro and LGastro). However, the MHam, LHam, AL, and Gr had slightly lower, but still good repeatability. Similarly, all tested muscles during running showed excellent between-days repeatability apart from TA and LHam, which still exhibited good to excellent repeatability. The reasons for the relatively lower reliability for these muscles probably could be its proximal origin (AL), acting on two different joints simultaneously (hamstring group), or variation in foot strike pattern (TA). All these factors could lead to a true variability while performing the dynamic tasks.

A number of factors could influence the reliability of EMG measurement in a clinical setting. These factors include extrinsic and intrinsic factors (Itoh et al., 2016). This includes several technical aspects such as the testing protocol, the measured muscles, the walking speed, body size, and the participant's age (Kuriki et al., 2012; Lyytinen et al., 2016). It is necessary to minimise the variation of EMG signals caused by these factors in order to improve the clinical value of EMG data (Burden & Bartlett, 1999). In the current study, high levels of repeatability of EMG was achieved in healthy subjects during walking and running. In order to enhance the repeatability of EMG measurement in this study, the aforementioned factors were controlled. For example, a rigorous protocol for locating the set of adductor muscles using ultrasonography was developed (see Table 3-2). In addition, proper skin preparation was undertaken during the testing sessions for all tested muscles. Another important factor which probably optimise the between days reliability is to have a standard testing protocol. In

this protocol, we standardised the walking and running speed and acceleration. Having such effective protocol results finally in having a good EMG data.

4.7.1.1 Comparison with previous EMG reliability study during walking

Studies using CMC values

The results of the tested muscles during walking revealed good to excellent CMC values (Table 4- 4, Table 4- 5, and Table 4- 6). Compared to previous studies, the CMC values in the results of this study appeared to have higher values. For example, the CMC values in the current study were higher than those reported by Kadaba et al. (1989) for GMax, GMed, AL, VMO, VLO, MHam, TA, and MGastro, and lower than the value in the LHam muscle. This may be due to the fact that in their study, EMG data were smoothed by a filter having a low-pass cut-off frequency of 12-14 Hz, compared to the smoothing used in the current study, which had an effective low (6 Hz) and high-pass (20 Hz) cut-off frequency for walking. Hence, the processed EMG data used here are inherently more variable compared to data used by Kadaba et al. (1989) for computing the CMC. This explanation is supported by Hershler and Milner (1978) and Kadaba et al. (1985) who found that the waveforms and repeatability results may be improved by providing an additional level of smoothing.

Studies using other reliability indices

The results of the current study are partly consistent with the findings of Kadaba et al. (1985), who showed that the repeatability of the EMG activity measured by surface electrodes is more consistent in all superficial muscles (VL, MHam and Gastro) except the TA muscle. This difference may be attributed to the fact that Kadaba et al. (1985) used the absolute value of VR to measure the reliability during free walking speed. However, in the current study, different repeatability indices were used. In addition, the participant was asked to walk and

run with a standard speed, and acceleration in order to optimise repeatable kinematics and kinetics parameters. Moreover, the results of the current study appeared to be quite similar to those of Murley et al. (2010) who reported good to excellent reliability for peak and RMS amplitude for the TA and MGastro muscles, while the corresponding values for absolute error were generally large in healthy young adults during walking at two self-selected speeds.

In the same context, Hubley-Kozey et al. (2013) reported good to excellent ICC values for EMG recordings in the VL, VM, MHam, LHam, LGastro, and MGastro muscles during walking at a self-selected gait speed. In addition, Lyytinen et al. (2016) concluded that the repeatability of the EMG activity in VM and LHam was good according to both the CV and ICC values in level walking. However, the measured EMG parameters, repeatability indices, and the study population were different. Only men were included in the current study, while Hubley-Kozey et al. (2013) investigated both men and women. In addition, some of the clinical characteristics of the subjects (e.g. BMI) differed between these studies. Thus, it is difficult to make a direct comparison between the current study and that of Hubley-Kozey et al. (2013).

SEM values

The current study shows that the SEM values for the AL, GR, and AM were higher than for the other muscles. This could be due to the dual roles of support and balance that result in true variability in movement performance in the proximal muscles and not a measurement error. In addition, they are primarily responsible for correcting the posture and balance of the dominant mass (2/3 of the body weight) of the head, arms and trunk. Such differences indicate that the adductor group has a somewhat more variable function than the other muscles, and this increased variability was seen during weight acceptance. Kadaba et al. (1985); Kadaba et al. (1989); Winter and Yack (1987) supported these results, finding greater

variability for proximal than for distal muscles, but no direct comparison can be made because of either the different variability index used or different walking speed between the studies' protocols. In addition, the inclusion criteria in the current study lead to the recruitment of a group of young healthy runners who might be considered to have highly repeatable motor control patterns and this could explain the higher levels of repeatability observed in this study, in comparison to other studies. In general, the present study revealed that the reproducibility of the EMG activity was not influenced by variation in gait speed but rather was dependent on the muscle studied and the normalisation method.

4.7.1.2 Consistency of EMG measurement in running

General discussion comparing studies using other reliability indices

The current study is the first study which has investigated the between day reliability of EMG measurements during running. The findings show good to excellent reliability for most of the measured muscles compared and these data compare well to the previous studies which have only determined the within day reliability (Table 4- 7, Table 4- 8, and Table 4- 9). In addition, this is the first attempt to investigate the reliability of EMG measurement for the hip adductors during running. For the hip extensor muscles, the results of the current study are similar to those of Smoliga et al. (2010), who studied the within day reliability of the EMG activity of the hip extensors during running. Their study was performed on a treadmill and the speed was gradually increased over five minutes until the participant reached 70% of maximal heart rate. Once this speed had been achieved, the participant was instructed to run for ten minutes. They reported that the reliability of EMG for the MHam and GMax were very good during running; however, the MHam was considerably more reliable and precise than the GMax. For the knee extensor muscles, the EMG activity of the VLO appeared to have low reliability, even less than the findings of the current study. This may be attributed to

using the different speed, different running surface (the current study was performed overground), different reliability index, and filtering used in their study. Such a different protocol could result in considerable variation in motor control of the entire knee joint, with the VL and VM activated to different degrees with each stride to stabilise the patella and maintain consistent varus/valgus angles. The discrepancy between these muscles may be related to differences in consistency of muscle activation patterns between muscles crossing one joint (GMax and VL) versus those that cross two joints (MHam and RF) (Prilutsky, Gregor, & Ryan, 1998).

CMC values for the adductor muscles

The current study revealed that the CMC values for the AL and GR were lower than for the other muscles. As previously mentioned when discussing the consistency of EMG measurement in walking, the EMG measurement for the proximal muscles is less reliable when compared to lower leg muscles. This could be due to the different roles played by the proximal muscles, as they support the stance limb and balance the upper body parts (head, arms, and trunks). For example, the adductor muscles have the capacity for generating force in the sagittal plane and therefore there is the possibility that the increased variability in this group was the result of differences in muscle synergy for generating the sagittal hip moments. For example, subjects may have used more hamstring/gluteal activity on one day and then utilised a higher activity of the AM on the second testing day. In line with this idea, it is possible that the observation of moderate reliability of the AL, Gr during running occur due to synergistic variation, i.e. differences in coordination pattern, rather than measurement error. Nevertheless, the results of the current study are partly consistent with the findings of Karamanidis et al. (2004) who found the same tendency during running. They found that the ICC values of the MGastro and LGastro were higher than those for VL, Ham, and TA.

Interestingly, three muscles of the same adductor group (AL, GR and AM) had different CMC values during running: AL was 0.74-0.78, GR was 0.68-0.74, and AM was 0.90-0.92. Such differences indicate that the AL and GR had a somewhat more variable function than the AM during running, and this increased the variability, which was seen during weight acceptance.

4.7.2 Effects of normalisation on between-session variability
4.7.2.1 Effect of normalisation technique on EMG amplitudes during walking

The second aim of the study was to compare the different normalisation techniques of muscle activation during walking and running. The best normalisation technique should result in minimal day-to-day variability but not require the removal of important information from the EMG signal. In general, the CMC values for the MVIC were only marginally lower than those observed using the MDT and PDT methods, with only a small number of significant differences across all the different muscle studied (Table 4- 10). Importantly, both the peak and the mean method involve normalising by some measure of amplitude in the processed gait signal and therefore require the removal of important, amplitude-related information from the EMG signal. Given that there was only small (mostly non-significant) differences in the reliability coefficient across the different methods of normalisation, and that both the PDT and MDT involve the removal of important information, it would seem appropriate to use the MVIC for normalisation of walking EMG signals.

Previous studies have found lower levels of repeatability using the MVIC method, in comparision to the PDT and MDT methods. For example, Burden et al. (2003); Shiavi, Bourne, and Holland (1986); Shiavi, Bugle, and Limbird (1987) and Yang and Winter (1984) found that the mean method resulted in slightly lower inter-individual variability than the peak method for most muscles analysed during normal gait. Shiavi et al. (1986); Shiavi et al. (1987) examined inter-individual variability for the GMed, RF, VLO, MHam, LHam, TA,

gastrocnemius, and soleus muscles and found similar or lower CV values for the mean than the peak during walking. Similarly, Burden et al. (2003) revealed that the inter-individual variability during walking was lower for mean and peak normalisation than for isometric and isokinetic normalisation methods for EMG activity of VL, VM, BF, and ST. Furthermore, Murley et al. (2010) found that the reliability of both TA and MGastro amplitude parameters was dependant on the normalisation techniques applied during overground walking. They found that relative reliability was moderate to good for MVIC normalised values, moderate for sub-maximum values, and very good to excellent for non-normalised values. However, a direct comparison with the findings of the current study cannot be made, as they were using different repeatability indices.

Both the mean and peak dynamic normalisation methods only serve to inform the researcher or clinician about the level of activity displayed by a muscle throughout the gait cycle in relation to the average and maximum activity, respectively, recorded during gait. In contrast, the MVIC method is designed to reveal how active a muscle is during gait in relation to its maximum capacity and therefore gives more insight into the overall level of activation than either of the mean or the peak method. As explained above, although the MVIC produced slightly lower repeatability compared to mean and peak dynamic normalisation methods, it provides useful information about the actual activity of the muscle in relation to its maximum capacity. Therefore, the data in this study support the use of the MVIC method in future EMG studies which are aimed at understanding differences between different individuals, for example when comparing healthy people and those with musculoskeletal condition.

Considering the MVIC techniques, the current study shows a relative lower reliability which could be attributed to that the MVICs may not represent the maximum activation capacity of the muscle in situations other than those at which the MVICs was performed. The

reproducibility of this reference point depends on subject`s level of sincerity, motivation. Such factors may result in MVIC variability and influence the interpretation of the EMG signal (Marras & Davis, 2001). Furthermore, the MVICs technique may increase activation variability by recruiting more type II muscle fibres than submaximal normalization techniques (Sinclair et al., 2015). In the current study, measurements were performed in a standardized manner. This included standardization of order of testing, rest time between each test, number of tests per muscle group, and verbal instruction and encouragement.

4.7.2.2 Effect of normalisation technique on EMG amplitudes during running

The current study is the first attempt to compare the reliability of methods used to normalise the EMG amplitudes during running. Similar to the walking task, the CMC values for the MVIC were also marginally lower than those observed using the MDT and PDT methods, with only a small number of significant differences across all the different muscle studied. The peak and mean normalisation methods do not utilise a reference contraction and therefore there is no physiological meaning to the magnitude of the data. As such, these approaches remove potentially important information on the signal magnitude that could differentiate between individuals with different kinematic/kinetic patterns. Given this potential limitation of the peak and the mean normalisation methods, the MVIC method was used to normalise the data in the following studies. However, it is important to note that, the MVIC normalisation method is influenced by the magnitude of the MVIC test and as such shows slightly lower repeatability. Nevertheless, the repeatability of the MVIC was still either moderate or excellent for the muscles tested and therefore appropriate to use for subsequent investigation.

The current study reliability results show that the difference between MVIC and PD are smaller than previous studies, which may be the result of the robust and standardized protocol

used in this thesis. Importantly, the data presented in this thesis shows good to excellent reliability using the MVIC method but also shows distinct variability in some muscles between different participants. The data also show that individual muscles have been approximately the same level of reliability value for the different methods of normalisation, e.g. VMO shows excellent (0.97-0.91), while the AL shows good reliability (0.75-0.83). Importantly, the results of the current study show small differences between the different methods of normalisation which are less than the other studies e.g. reliability data has been reported for VM of 0.77, 0.92, 0.61 for the MDT, PDT and MVIC respectively (Sinclair et al. 2012). Given the similarity of the reliability indices for the three different techniques, it would seem appropriate to use the MVIC method if a physiologically meaningful interpretation of muscle activation is required.

The relatively lower MVIC reliability may be due to the variability of MVIC between the two testing days. Similar to the walking task, the AL represents the least CMC value and the GMax shows the highest value. The reason could be the higher level of variability and lower reliability while generating the MVIC itself. This was reinforced by the results of the between days reliability for MVIC values (Table 4- 13). The ICC results show that few muscles (AL and VLO) that produce poor level of reliability while performing MVIC, which in turn resulted in low CMC value. On contrary, the muscles that have excellent between days reliability (e.g. GMax, AM) displayed excellent CMC value. Consequently, the consistent MVIC measurement reduces the variations among the different normalisation techniques. In addition, the muscles that show difficulty in producing consistent MVIC values across testing days will display higher level of between days variability (less CMC value).

The relatively lower reliability of MVICs could be attributed to the fact that the MVICs may not represent the maximum activation capacity of the muscle. The reproducibility of this

reference depends on subject`s level of willingness to engage and motivation. Such factors may result in MVIC variability and influence the interpretation of the EMG signal (Marras & Davis, 2001). Furthermore, the MVICs technique may increase activation variability by recruiting more type II muscle fibres than submaximal normalization techniques (Sinclair et al., 2015). In the current study, measurements were performed in a standardized manner. This included standardization of order of testing, rest time between each test, number of tests per muscle group, visual feedback and verbal instruction and encouragement. By using such standardisation, every attempt was made to control, as far as possible, for possible inaccuracy in the MVIC measurements. The good to excellent reliability of the MVIC contractions for most of the muscles studied, shown in Table 4- 12, shows that our robust protocol led to consistent MVIC data and supports the subsequent use of MVIC data to normalise the dynamic EMG data.

Previous studies (Albertus-Kajee et al., 2011; Ball & Scurr, 2011; Sinclair et al., 2012) investigated the reliability of different normalising techniques during running. However, direct comparison cannot be made with the current results as the previous studies focus on the approach which reduce the between subject variability and the within day rather the between days reliability or using different normalisation techniques. In conclusion, it appears that different normalisation techniques should not be used interchangeably in the analysis of EMG data. This may facilitate misinterpretation of the EMG amplitude and using the most appropriate normalisation technique is critical to achieve clinically reasonable findings. In addition, MVIC provided slightly lower repeatability, but it affords useful information about the motor control pattern during dynamic task. Therefore, it is recommended to be used especially when the variation between subjects in musculoskeletal disorder is explored.

4.7.3 Reliability of kinematics measurements

The third aim was to assess the between-day reliability of measuring 3D biomechanical variables during walking and running activities. The results show excellent reliability of all measured angles during walking and running tasks (Table 4- 14 and Table 4- 15). The sagittal plane angles such as hip and knee angles displayed slightly higher CMC values than the frontal plane angles.

4.7.3.1 Reliability of kinematics measurements during walking

The results show that when subjects were instructed to walk at a constant speed, they tended to keep consistent hip and knee sagittal plane movements compared to slightly more variable hip and pelvis frontal plane movements. For frontal plane angles, the between-day repeatability was excellent, demonstrating that the marker reapplication error was minimal. The relative lower repeatability of the frontal pelvic movement may be due to its small range (peak value ≈ 6-9º), as well as a lack of a well-defined displacement pattern. Consequently, the between day reliability for the frontal plane kinematics was slightly affected. In general, the slightly lower repeatability for pelvis and hip joint motion in the frontal plane indicates that the variability in pattern of walking was reflected mostly in the frontal plane motions.

The results of the current study concur with those reported by Kadaba et al. (1989), who found that in the sagittal plane, the repeatability of joint angle motion at the hip, knee, and ankle were excellent while the pelvic tilt pattern displayed the lowest repeatability during normal walking. In addition, the CMCs for the joint angle motion in the frontal and transverse planes were lower than those for sagittal plane motion. The findings from Tsushima et al. (2003) also support the current study results, as the authors found excellent reliability for hip and knee motions in the sagittal plane and pelvic tilt in the frontal plane. Moreover, the current study results appear to be quite similar to those of Growney et al.

(1997), who assessed the repeatability of sagittal, frontal, and transverse plane angles for hip, knee, and pelvic tilt while walking at self-selected speeds. They found that the sagittal plane angles for the hip and knee demonstrated excellent repeatability, while the frontal plane angles were fairly repeatable.

4.7.3.2 Reliability of kinematics measurements during running

Similar to the walking task, the reliability of kinematic data during running showed that the CMC value in the sagittal plane was the highest value compared to relatively low values for the hip and pelvis in the frontal plane. To some extent, the current study results were similar to findings from Mason et al. (2016). They obtained high CMC values for the sagittal plane motion of the hip and knee and slightly lower CMC values for the frontal pelvic movement during running. Interestingly, the findings of the current study demonstrated good reproducibility for frontal pelvic movement (high CMC and lower SEM values) compared to the results of Mason et al. (2016). This difference could have resulted from different speeds used (5.6 vs 3.2 m.s^{-1}) or the errors in repositioning either the ASIS or the PSIS markers used to detect the pelvic frame in the previous study. In the current study, a number of training sessions were performed by the rater before the actual data collection, thus optimising the pelvic marker placement. Similar to the findings of the current study, Schache et al. (2002) investigated the kinematic reproducibility of the hip and pelvis during running. They reported excellent CMC values of sagittal and frontal hip motions (0.972, 0.989 respectively) that are similar to those found in the current study. However, poor between-day repeatability was found for the frontal pelvic motion. This difference may be attributed to the better marker placement in the current study.

In the same context, the SEM values in our study were somewhat lower compared to previous studies. Noehren et al. (2010) reported SEM values for sagittal motion of the hip (5.1º) and

knee (1.9º) angles, suggesting better reproducibility in our study. This difference may be attributed to the slightly lower speed used in our study, which may have resulted in relatively low levels of between-day variability. In the same manner, the results of Ferber et al. (2002) showed low ICC and higher SEM scores for the sagittal plane motion of the hip and knee joint angles during running. They found that the ICC values for the hip and knee were 0.88 and 0.93 respectively, and the SEM values were relatively higher (2.21º and 2.22º respectively). It was noticed that the differences may be attributed to individual subject differences (both genders vs males only in the current study), as well as the variations in running speed (3.46-3.83 vs 3.2 m.s^{-1}), which resulted in higher variability of the kinematic gait patterns (van der Linden, Kerr, Hazlewood, Hillman, & Robb, 2002). Significant changes in speed across sessions are therefore more likely to be associated with 'true' change in kinematic variability, rather than error related to inconsistent marker placement (McGinley et al., 2009). In summary, data of the current study showed excellent CMC values between testing days, and this was depicted by the very low values of the SEM for the kinematic variables during both tasks, which provides confidence in the proposed testing protocol.

4.8 Limitation

It was difficult to ensure the maximum effort of each participant during the measurement of the MVIC. A number of methods were tried in this study to improve the quality of the MVIC measurements. This included a detailed explanation of the importance of giving the maximum effort during measurement and how this could influence the interpretation of the results. In addition, consistent verbal encouragement and visual feedback were provided during the measurement of the isometric contraction. A consistent testing position, as recommended by SENIAM, was also used between the testing days along with a standard testing order, consistent rest time (minimum of 30 sec) between each test and a standard testing number for each muscle group. However, despite these measures, the MVIC will still

introduce an additional source of variability in the between session measurements. This may be due to the fact that some individuals are able to activate their muscles closer to their true maximum than others and because of between-session variability in. maximal exertions. Nevertheless, most of the muscle's studies showed relatively high ICCs for the between-day MVIC data (Table 4-12) & 4-13). Given that this measure of consistency will capture variability in the maximal EMG signal due to the removal and subsequent replacement of electrodes, the true repeatability of the MVIC data is likely to be higher. These data therefore suggest that using MVIC data to normalise dynamic EMG signals is an appropriate and valid approach, especially as it does not lead to the removal of important amplitude information from the dynamic signals. Previous authors (Allison et al., 1993; Knutson et al., 1994), warned against using normalisation methods that reduce the true variation of EMG patterns between individuals. Consequently, it is suitable for use MVIC technique in the subsequent chapters. Specifically, the combination of relatively high CMCs (0.70–0.96) and lower SEMs (0.02 - 0.30) associated with data expressed as a % MVIC demonstrated moderate to excellent measurement reliability and stability.

4.9 Conclusion

The most important finding of this repeatability study is that the EMG activity exhibited good to excellent between-day repeatability in the majority of the tested muscles during both walking and running activities. Considering the method of normalisation, EMG amplitudes normalised to the peak and mean of the dynamic trials demonstrated a high CMC and low SEM. Importantly, although the MVIC normalisation lead to slightly lower repeatability indices, it provides useful information about the actual activity of the muscle in relation to its maximum capacity. In the final chapter, there was a strong focus on exploring the extent to which inter-subject variability in EMG patterns could explain variation in pelvic kinematics. Therefore, given that the PDT and MDT remove potentially important information from the

EMG signal and that reliability is only slightly higher, the MVIC method was deemed to be the most suitable method. For the kinematic variables, the repeatability of these variables was lower in running compared to walking, either in the sagittal or frontal planes. Moreover, the sagittal plane movement for hip and knee joints showed excellent reliability both in walking and running. However, although the frontal pelvic movement showed a slightly lower repeatability, it was still observed to have excellent between days reliability.

Chapter 5: EMG profile for lower limb muscles during running.

5.1 Introduction

In the previous two chapters, a standard and reliable technique for measuring the activity of the adductor muscles during walking and running using surface EMG electrodes was developed. In Chapter 3, a robust method to measure the activity of the adductors during dynamic tasks was established. This method provides a high confidence, as it minimised the possibility of cross-talk from the adjacent muscles. In addition, the surface EMG electrodes placed over the adductor muscles in this method were sensitive to minor changes in muscle output. Moreover, using same method, the adductor muscles exhibited good to excellent between-days reliability during stance phase of both walking and running activities (see Chapter 4). This particular phase was selected because the maximum degree of pelvic drop (the period of interest in the next study, Chapter 6) occurs during early stance. In addition, the muscles supporting the pelvis at this period are activated around this phase. For example, the highest peak of AM activity occurs early in stance. Moreover, analyses were restricted to the stance phase of gait as a previous study indicated that CMCs can be falsely inflated due to muscle inactivity during the swing phase (Barn et al., 2012). Having a good method for measuring the activity of the adductor muscles, the EMG profile for their activity during the stance phase of running will be described. From this chapter onwards, the focus will be on understanding the role of the adductor muscles in running activity.

Previous studies describing the EMG profile for the adductor muscles

A minimal number of studies describe the EMG profiles for the hip adductor muscles during running. Such profiles were measured using either surface or fine wire electrodes.

Montgomery et al. (1994) proposed three peaks of activity for the AM during running; the highest peak occurs early in stance during the loading response, the second peak during the early swing, and the third peak in middle swing. The three peaks occur in late swing at 90%, in midstance at 18%, and in midswing at 68%. These peaks become prominent only at 3 m.s^{-1} and higher, at lower running speeds the EMG is low and irregular, even to the extent that it is hard to see periodicity (Gazendam & Hof, 2007). In contrast, the AL activates around the toe off and continues in the early swing (Chaudhari et al., 2014).

The aforementioned studies revealed a high level of variability among researchers in their understanding of how the adductor muscles activate during running activity. The reason behind this variability could be a lack of standardisation in the method used to record the activity of these muscles. It could also be due to the use of two different recording EMG electrodes (fine wire and surface electrodes). Another possible reason could be true variability between individuals when performing the running task. Thus, with a standard and reliable recording method using surface electrodes (see chapters 3 and 4), it may be possible to gain a better understanding of how this group of muscles work during gait.

Factors affect the EMG profile during running

A number of measurement factors are proposed to have a direct influence on muscle activation pattern during running. These factors include the running surface, the running speed, the running acceleration, the degree of inclination, and the strike pattern. The running surface could significantly impact on the muscular activity while running. For example, Wank et al. (1998) reported that the biceps femoris and the RF showed a higher magnitude during treadmill running, while the VLO showed lower amplitudes. Similar to the running surface, the change in the running speed is proposed to have an influence on the muscle activation profile. The higher the speed of running, the higher the activity of the lower limb

muscles (Jensen, Leissring, & Stephenson, 2016; Wall-Scheffler et al., 2010; Yokozawa, Fujii, & Ae, 2007). Additionally, Chumanov, Wille, Michalski, and Heiderscheit (2012) proposed that the effect of increasing speed on muscle activity differs according to the phase of the gait cycle. In the same way, It has been proposed that fluctuation in running acceleration could greatly influence the EMG activation pattern (Frost et al., 2010; Sakamoto & Sinclair, 2012; Tokuda et al., 2016).

Similarly, moving from a level to an inclined surface is thought to have an influence on muscle activation pattern (Lay, Hass, Nichols, & Gregor, 2007; Swanson & Caldwell, 2000). For example, Wall-Scheffler et al. (2010) found that running up a slope, the biceps femoris, VLO, GMed, and GMax showed a significant increase in their electromyographic amplitudes. Similarly, hip adductors displayed increased activity across the gait cycle when increasing the running slope. For the strike pattern, Lieberman et al. (2010) proposed that the change in foot strike pattern could influence the electromyographic activity and the kinematics of the lower limb (Lieberman et al., 2010). For example, a considerable increase in the EMG amplitudes of the gastrocnemius at initial contact was measured in rearfoot strikers (RFS) compared to its pre-activation level (about four times) (Shih, Lin, & Shiang, 2013). Likewise, Ahn, Brayton, Bhatia, and Martin (2014) found that forefoot strikers activated their plantar flexor muscles 11% earlier and 10% longer than RFS early in stance. The aforementioned factors that could affect the EMG profile must be considered while establishing the testing protocol.

The study presented in this chapter aimed to describe the EMG profile for the adductor muscles and their variability during running. Additionally, it is aimed at describing another nine lower limb muscles as well as their associated variability during the running. In order to decrease the variability between subjects, a robust testing protocol was developed. This

protocol standardises the running speed and the running acceleration, running shoes and the running surface.

5.2 Aims

This study aimed to describe the typical EMG profile for the adductor muscles during running as well as their inter-subject variability. In addition, the study aimed to describe the typical EMG profile for the major lower limb muscles collected during running as well as the associated inter-subject variability.

5.3 Methodology
5.3.1 Recruitment plan

Similar to the recruitment plan applied in chapter 4, individuals were invited to participate in this study through a number of avenues (see Chapter 4, section 4.3.1). This included placing posters around the university campus, emailing the local running/triathlon clubs, and advertising via the running performance clinic website. A number of responses from many runners was received, and only those who met the entry criteria were selected for participation in this study.

5.3.2 The entry criteria

The same inclusion criteria applied in the previous chapter were applied in this study as well (see Chapter 4, section 4.3.2). All participants were males aged between 18-40 years. The participant needed to be free from lower-limb injuries in the past six months. In addition, the training routines for all participants had to include running training at least three times per week for a minimum of 10-15 miles per week. This training programme was performed for at least three months prior to enrolment in the study.

5.3.3 Participants

A cohort of 25 male runners, with no history of lower-limb injury or surgery, was recruited for this study. The mean (SD) age of the subjects was 32 (8.1) years, mean (SD) height was 1.79 (0.1) m, mean (SD) weight was 73.2 (10.40) kg, and the mean (SD) body mass index (BMI) was 22.76 (2.1) kg/m^2. The data obtained from the 25 participants was used to describe the EMG profile for the AL, Gr, AM, GMax, GMed, VMO, and MHam. In addition, 10 more participants from the repeatability study were added to the current study in order to describe 5 more muscles, VLO, LHam, MGas, LGas, and TA. The demographic data of the later 10 participants were presented in chapter 4, section 4.3.3. Only 4 participants were accepted to participate in the different studies conducted throughout the whole thesis.

5.3.4 Ethical approval

Before starting the data collection, all participants read and signed a written informed consent statement approved by the Research, Innovation, and Academic Engagement Ethical Approval Panel at the University of Salford (Appendix IV).

5.3.5 Instrumentation

The same equipment used in the previous studies was used here. This included an ultrasound imaging system (MyLab70, Esaote, USA), a Direct Transmission System with 16 channels (Noraxon USA inc., model 586 Tele Myo DTS Desk Receiver), the DTS sensors (model 542), EMG lead (542AP), and a disposable adhesive Ag/AgCl EMG electrode (for more details, see the methodology section in Chapter 3). Additionally, a motion capture system (ten Pro-Reflex, Qualisys cameras with three embedded force platforms) was used in this study in order to identify the gait events (for more details, see the methodology section in chapter 4). The set-up procedures for the ultrasound, EMG measurements, and motion capture system

used in the previous studies were followed here as well (see Chapter 3, sections 3.4.3 and 3.4.5; and Chapter 4, section 4.3.5).

5.3.6 Surface EMG electrode and 3D marker placement

After signing the consent form, the participants were asked to change into their shorts and a comfortable t-shirt. Then, the height and weight of the participants were measured. This was followed by *placement of the surface EMG electrodes*. Similar to Chapter 4, the muscles were located according to the SENIAM guidelines (Hermens et al., 1999). The EMG amplitudes were obtained from seven muscles: GMax, GMed, VMO, AL, Gr, AM, and MHam. The adductor group (AL, Gr, and AM) was located using similar approaching techniques to those used in study 1 (see Chapter 3, table 3-2). Finally, all steps for skin preparation and the signal-checking process followed in Chapter 3 were followed here (see Chapter 3, section 3.4.5).

Marker placement. 40 reflective markers of 14.5 mm diameter were used in all trials of data collection. The markers were attached to the skin using hypoallergenic double-adhesive tape attached to a flat-based marker. The marker set was distributed over the different lower limb parts in a similar way to the previous study (see Chapter 4; section 4.3.6).

5.3.7 Testing protocol

The procedures of the testing protocol undertaken in the previous chapter were followed in this chapter as well (see Chapter 4, section 4.3.7). This included the protocol for recording the static and dynamic trials, in addition to the protocol for obtaining the MVIC. The adductor muscles were normalised to the MVIC according to the results of the pilot study undertaken in Chapter 4 (see Chapter 4, table 4.1).

5.4 Data processing

EMG processing. The processing plan of the EMG data applied in the previous study was followed here in this study (see Chapter 4, section 4.4.1). This included removal of movement artefacts, averaging the gait data and time normalisation. Moreover, the MVIC was processed in a way similar to the previous study (see Chapter 4, section 4.4.1). The data presented in this chapter were normalised to both MVIC of PDT across the stance phase of running for all subjects n=25 or n= 10 depending on the tested muscle. Finally, the coefficient of variance (CV) was calculated by dividing the standard deviation by its mean value (Winter & Yack, 1987). As this study is the first attempt to understand the variability of the adductor muscles during running, the CV was calculated at three different points in the stance phase, 0, 50, and 100% respectively. The reason for choosing these points in the stance phase is to provide an overview of the level of the variability of the different muscles over the stance phase of running. Moreover, the majority of the muscles of interest related to pelvic drop, studied in the final study, peaks around the initial contact. In addition, the inter-subject variability of the stance time of the running cycle was calculated.

In next section (the results), the profile of the adductor group will be presented. This includes the EMG profile for the AL, Gr, and AM muscles plus the profile for the major lower limb muscles (GMax, GMed, VMO, VLO, TA, MHam, LHam, MGastro, and LGastro). The scale represents the stance phase and pre-activation period (the minus sign refers to the activation period before foot contact).

5.5 Results

5.5.1 Adductor group profile

In this section, the profile of the adductor muscle group will be described. This includes the EMG profile for the AL, Gr, and AM muscles. The scale represents the stance phase and pre-

activation period (the minus sign refers to the activation period before foot contact). Table 5-1 shows the average (SD) of the EMG amplitudes for the adductor muscles over the stance period. In addition, it summaries the CVs for the tested muscles. For each muscle, the data were displayed at three points of the stance phase, 0, 50, 100%, using two different normalisation techniques, MVIC and PDT. Generally, the inter-subject variability was quite large in all tested muscles when normalised to MVIC. When the ensembles were normalised to the PDT, the inter-subject variability reduced in all muscles.

Adductor longus

Regardless of the method of normalisation, AL activated at about 25% of the stance phase and continued throughout the stance phase (Figure 5- 1). It displayed one peak at around 75% of the stance. The peak reached about 40% of MVIC when normalised to MVIC, while the peak reached about 80% of PDT when normalised to PDT. The AL ensemble normalised to PDT provided the lowest inter-subject variability compared to the AL ensemble normalised to MVIC. When the EMG amplitudes was normalised to PDT, lower inter-subject variability was observed at 0% of the stance phase, while for the amplitudes normalised to MVIC, lower variability was observed at 100% of the stance phase.

Figure 5- 1: The figure displays the ensemble average (± 1 SD) of: a. adductor longus (AL) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. AL when normalised to the peak activity of the dynamic trials (N-PDT). 25 participants were used to create the profile of this muscle.

Gracilis

The gracilis showed a profile starting from the pre-activation period and ending at about 75% of the stance phase, regardless of the method of normalisation (Figure 5- 2). It showed one peak at around the 50% of the stance. When the Gr amplitudes were normalised to MVIC, the peak reached about 40% of MVIC, while the amplitude normalised to PDT reached about 80% of PDT. The Gr ensemble normalised to PDT provided the lowest inter-subject variability compared to the ensemble normalised to MVIC. The lower inter-subject variability was seen at 100% of the stance phase for both normalisation techniques.

Figure 5- 2: The figure displays the ensemble average (± 1 SD) of: a. gracilis (Gr) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. Gr when normalised to the peak activity of the dynamic trials (N-PDT). 25 participants were used to create the profile of this muscle.

Adductor magnus

Adductor magnus demonstrated a profile starting from the pre-activation period and ending at about 75% of the stance, regardless of the method of normalisation (Figure 5- 3). It showed one peak at about 25%. When the AM amplitudes were normalised to MVIC or PDT, the peak reached about 70% of MVIC or 70% of PDT; respectively. The AM ensemble normalised to PDT provided the lowest inter-subject variability compared to the ensemble normalised to MVIC. When the AM amplitudes were normalised to PDT, lower inter-subject variability was seen at 100% of stance phase, while for amplitudes normalised to MVIC, lower variability was observed at 0% of the stance phase.

Figure 5- 3: The figure displays the ensemble average (± 1 SD) of: a. adductor magnus (AM) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. AM when normalised to the peak activity of the dynamic trials (N-PDT). 25 participants were used to create the profile of this muscle.

Table 5- 1: Inter-subject EMG variability of adductor muscles during the stance phase of running. The values were presented as average amplitudes (SD) normalised to maximum isometric voluntary contraction (%MVIC) and peak of the dynamic trials (%PDT). The values were presented at three different points in the stance phase (SP), 0, 50, and 100% respectively.

	No.	% SP	N. MVIC			N. PDT		
			AV. Amplitudes (%MVIC)	SD	CV (%)	AV. Amplitudes (% PDT)	SD	CV (%)
AL	25	0%	12%	6%	48	20%	6%	28
		50%	36%	34%	92	57%	23%	40
		100%	33%	13%	39	69%	25%	36
Gr	25	0%	15%	8%	56	34%	19%	55
		50%	39%	35%	89	68%	29%	43
		100%	12%	6%	52	26%	10%	40
AM	25	0%	49%	27%	56	55%	22%	39
		50%	49%	41%	84	52%	28%	53
		100%	12%	8%	67	15%	6%	37

5.5.2 The EMG profiles of the major lower limb muscles during the stance phase of running

In this section, the EMG profiles for nine lower limb muscles will be described. In order to describe each EMG profile, the muscles are grouped into a number of functional groups: (1) a gluteal group that involves the EMG profile for GMax and GMed; (2) a quadriceps group that includes the EMG profile for VMO and VLO; (3) a hamstring group that contains the EMG profile for MHam and LHam; (4) a gastrocnemius group that contains the EMG profile for MGastro and LGastro; and (5) the EMG profile for the TA. The scale represents the stance phase and pre-activation period (the minus sign refers to the activation period before foot contact).

Table 5- 2 shows the average (SD) of the EMG amplitudes for the nine lower limb muscles over the stance period. In addition, it summaries the CVs for the tested muscles. For each muscle, the data were displayed at three points of the stance phase (0, 50, 100%) using two different normalisation techniques, MVIC and PDT. Generally, the inter-subject variability was quite large in all tested muscles when normalised to MVIC. When the ensembles were normalised to PDT, the inter-subject variability reduced in almost all muscles. With PDT normalisation technique, lower inter-subject variability was seen at 0% of the stance phase for all muscles except for the MGastro and LGastro muscles (their lower variability occurred at 50% of the stance phase). When EMG activity was normalised to MVIC, lower inter-subject variability was seen at 0% of the stance phase for all muscles except for the VMO and LGastro muscles (their lower variability was observed at 100% and 50% of the stance phase, respectively). The mean (SD) stance time of the running gait was 0.26 (0.03) ms and the inter-subject variability of the stance time was 10.74%.

Table 5- 2: Inter-subject EMG variability of the major lower limb muscles during the stance phase of running. The values were presented as average amplitudes (SD) normalised to maximum isometric voluntary contraction (%MVIC) and peak of the dynamic trials (%PDT). The values were presented at three different points in the stance phase (SP), 0, 50, and 100% respectively.

	No.	% SP	N. MVIC			N. PDT		
			AV. Amplitudes (%MVIC)	SD	CV (%)	AV. Amplitudes (%PDT)	SD	CV (%)
GMax	25	0%	28%	12%	43	75%	20%	27
		50%	13%	10%	79	34%	15%	42
		100%	4%	4%	102	11%	5%	42
GMed	25	0%	48%	20%	42	65%	15%	24
		50%	20%	15%	74	27%	18%	65
		100%	9%	6%	64	12%	7%	60
VMO	25	0%	68%	31%	45	62%	17%	27
		50%	40%	20	51	36%	14%	38
		100%	7%	2%	33	6%	3%	54
VLO	10	0%	69%	25%	37	55%	12%	22
		50%	42%	28%	67	34%	16%	46
		100%	5%	2%	40	4%	1%	24
TA	10	0%	37%	13%	34	71%	16%	23
		50%	20%	16%	82	36%	25%	70
		100%	16%	8%	54	30%	17%	56
MHam	25	0%	17%	8%	50	35%	17%	49
		50%	18%	13	72	40%	25%	62
		100%	03%	3	77	8%	5%	60
LHam	10	0%	16%	6%	39	36%	13%	37
		50%	33%	22%	67	72%	29%	40
		100%	4%	2%	68	9%	8%	80
MGastro	25	0%	28%	10%	36	28%	15%	54
		50%	96%	31%	32	93%	7%	8
		100%	2%	1%	43	3%	2%	89
LGastro	25	0%	23%	9%	39	26%	9%	33
		50%	78%	24%	31	90%	10%	12
		100%	4%	4%	122	4%	5%	112

5.5.2.1 Gluteal group

Regardless of the method of normalisation, both glutei showed nearly the same profile.

GMax activity started before foot contact (about -20% pre-stance) and ended at about 60% of

the stance phase (Figure 5- 4). GMed activated just after GMax (about -15% pre-stance) and

ended at about 55% of the stance phase (Figure 5- 5). Both glutei showed a profile with one

peak at 15% of the stance phase. When the GMax amplitudes were normalised to MVIC, the peak reached about 35% of MVIC and 90% of PDT when normalised to PDT. When the GMed amplitudes were normalised to MVIC, the peak reached about 70% of MVIC and 90% of PDT when normalised to PDT. The GMax and GMed ensemble normalised to PDT provided the lowest inter-subject variability compared to the ensemble normalised to MVIC. Lower inter-subject variability was seen at 0% of the stance phase in both normalisation techniques.
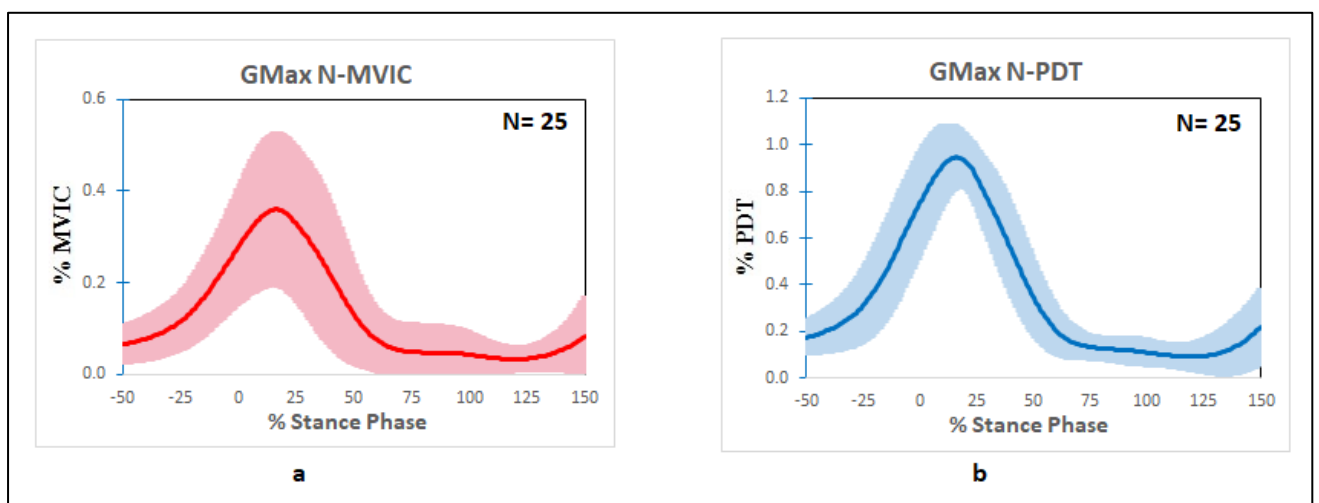
Figure 5- 4: The figure displays the ensemble average (± 1 SD) of: a. gluteus maximus (GMax) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. GMax when normalised to the peak activity of the dynamic trials (N-PDT). 25 participants were used to create the profile of this muscle.
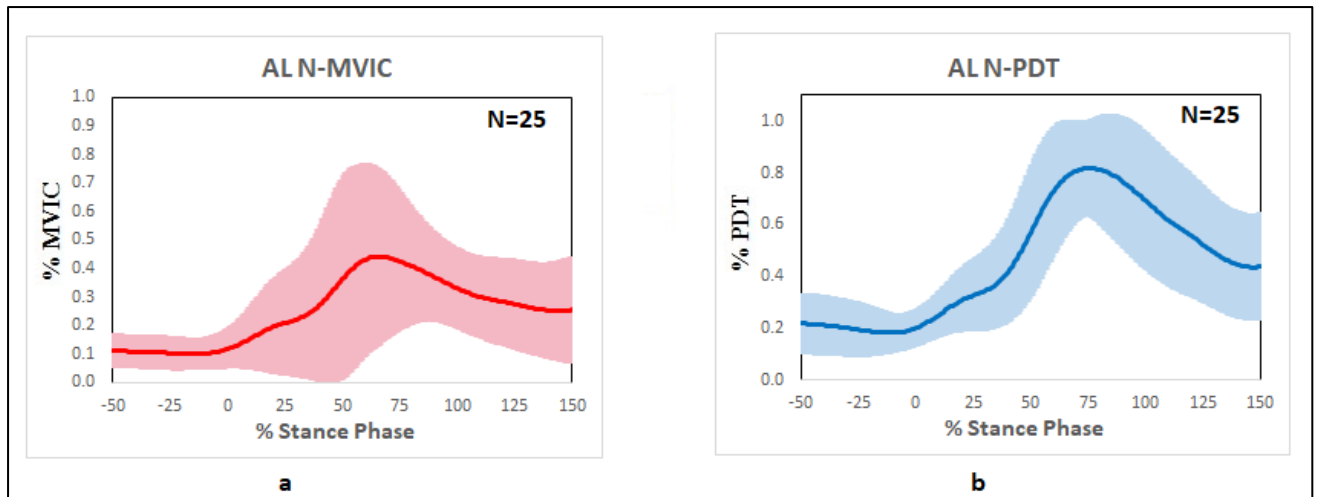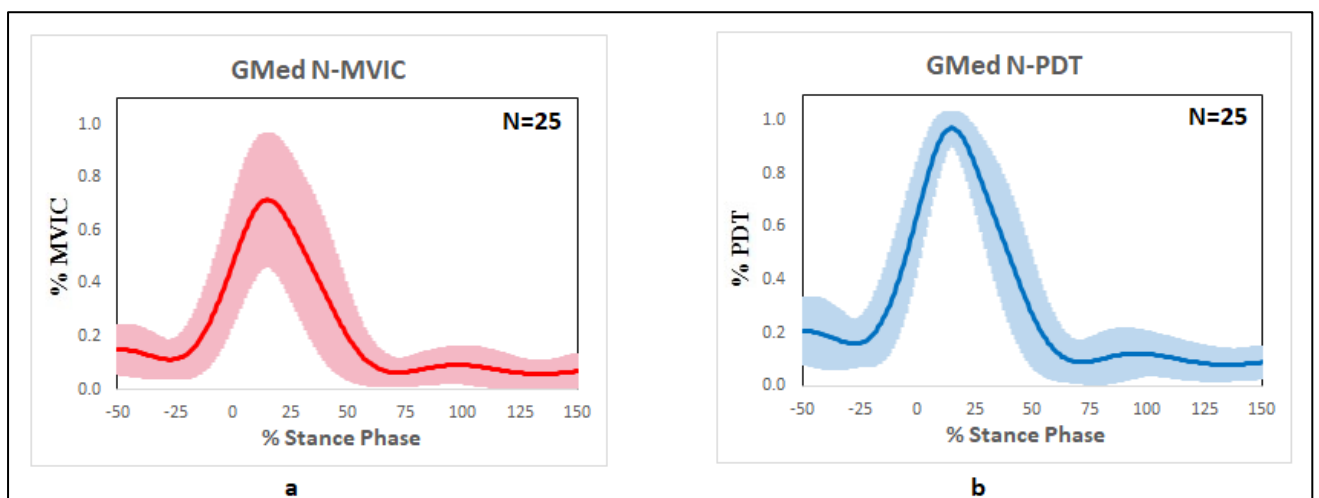

Figure 5- 5: The figure displays the ensemble average (± 1 SD) of: a. gluteus medius (GMed) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. GMed when normalised to the peak activity of the dynamic trials (N-PDT). 25 participants were used to create the profile of this muscle.

5.5.2.2 Quadriceps group

Apart from the normalisation method, the profile for VMO and VLO started before foot contact (about -20% pre-stance) and ended at about the midstance (about 60% of the stance), with one peak at around the 20% of stance (Figure 5- 6 and Figure 5- 7). Twenty-five participants were used to create the profile for VMO, while the profile for VLO was established based upon the EMG data of ten participants. When EMG amplitudes of VMO and VLO were normalised to MVIC, the peak reached about 100% and 110% of MVIC, respectively. The peak of VMO and VLO reached 95% of PDT when normalised to PDT. Ensembles of VMO and VLO normalised to PDT provided the lowest inter-subject variability compared to ensembles normalised to MVIC. When the amplitudes were normalised to MVIC, VMO displayed lower inter-subject variability at the 100% of stance, while VLO displayed lower inter-subject variability at 0% of stance. When normalised to PDT, both vasti demonstrated lower inter-subject variability at 0% of the stance.
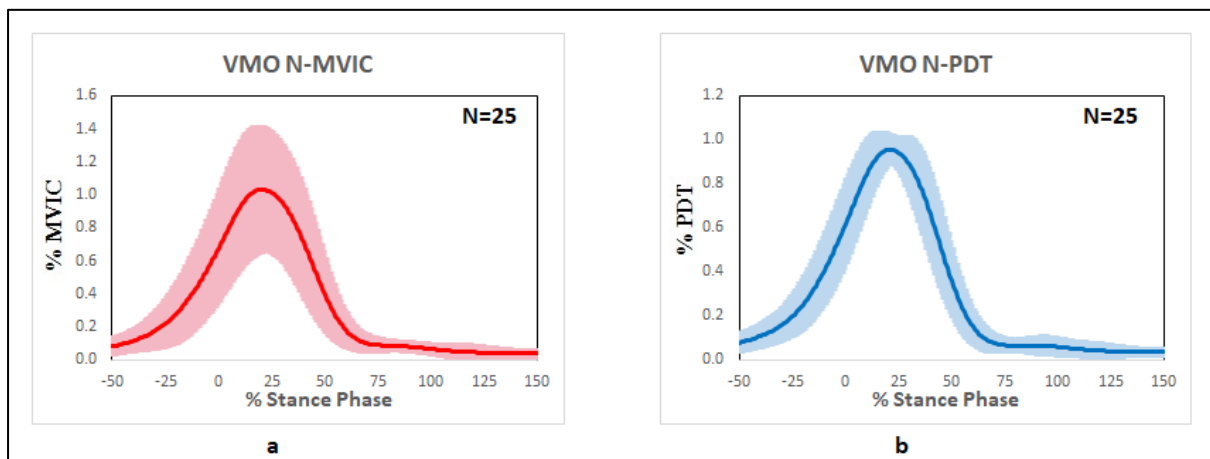
Figure 5- 6: The figure displays the ensemble average (± 1 SD) of: a. vastus medialis obliques (VMO) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. VMO when normalised to the peak activity of the dynamic trials (N-PDT). 25 participants were used to create the profile of this muscle.
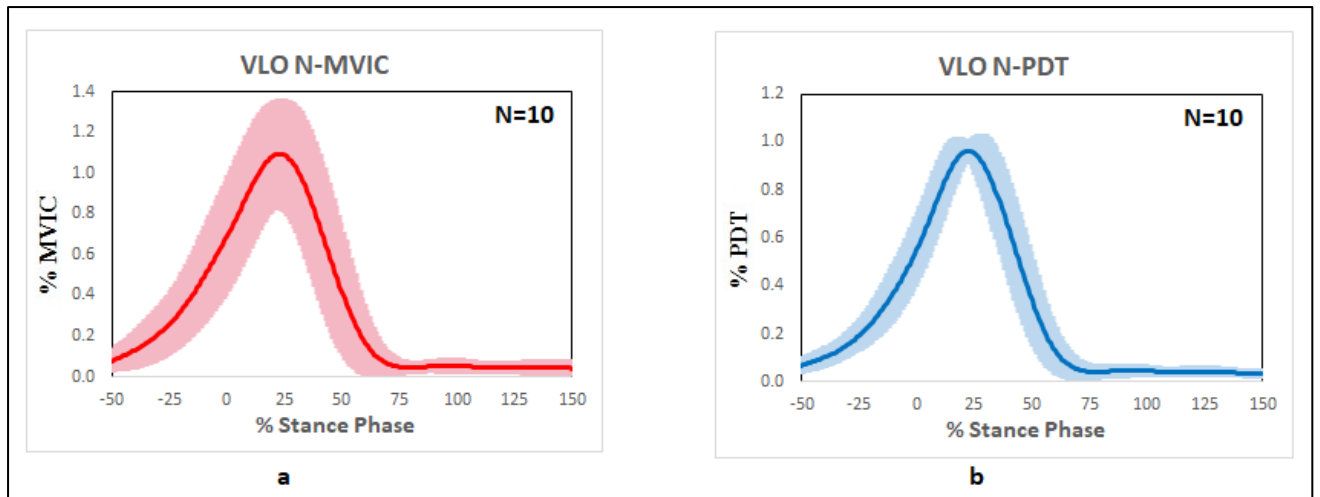
Figure 5- 7: The figure displays the ensemble average (± 1 SD) of: a. vastus lateralis obliques (VLO) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. VLO when normalised to the peak activity of the dynamic trials (N-PDT). Ten participants were used to create the profile of this muscle.

5.5.2.3 Hamstring group

For both normalisation methods, the hamstring group profiles showed activity started at the pre-stance and continued approximately up to 75% of the stance phase. The profile for both hamstrings muscles showed two peaks: the highest occurred at the pre-stance and the second peak occurred at around 50% of the stance phase (Figure 5- 8 and Figure 5- 9). When the amplitudes of MHam and LHam were normalised to MVIC, the peak reached about 20% and 30% of MVIC, respectively. In contrast, the peaks of MHam and LHam reached 40% and 70% of PDT when normalised to PDT, respectively. Ensembles of MHam and LHam normalised to PDT provided the lowest inter-subject variability compared to ensembles normalised to MVIC. For both hamstrings, lower inter-subject variability was seen at 0% of the stance phase for both normalisation methods.
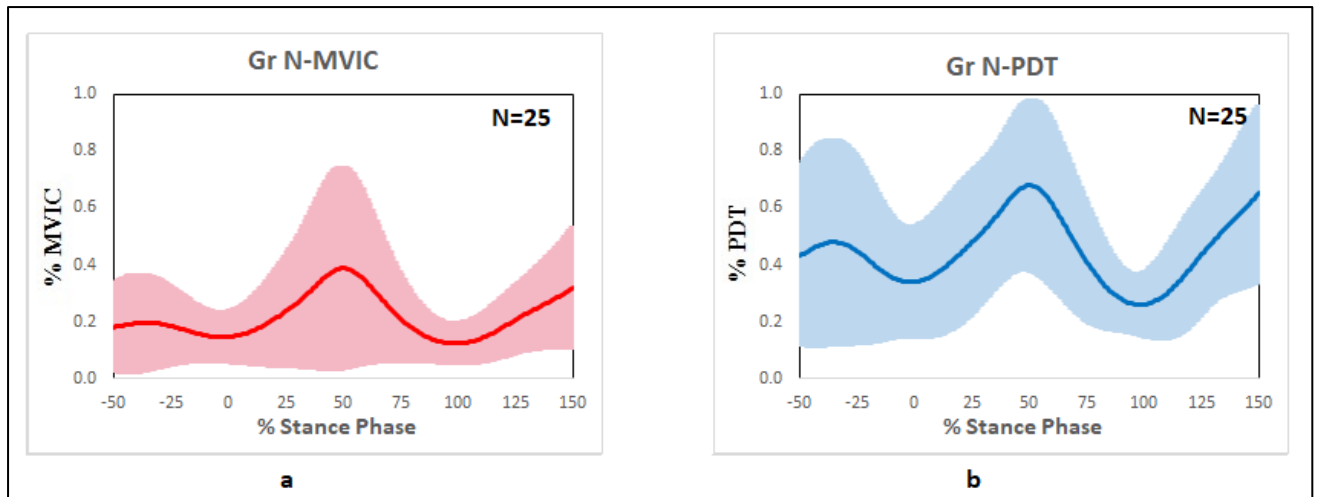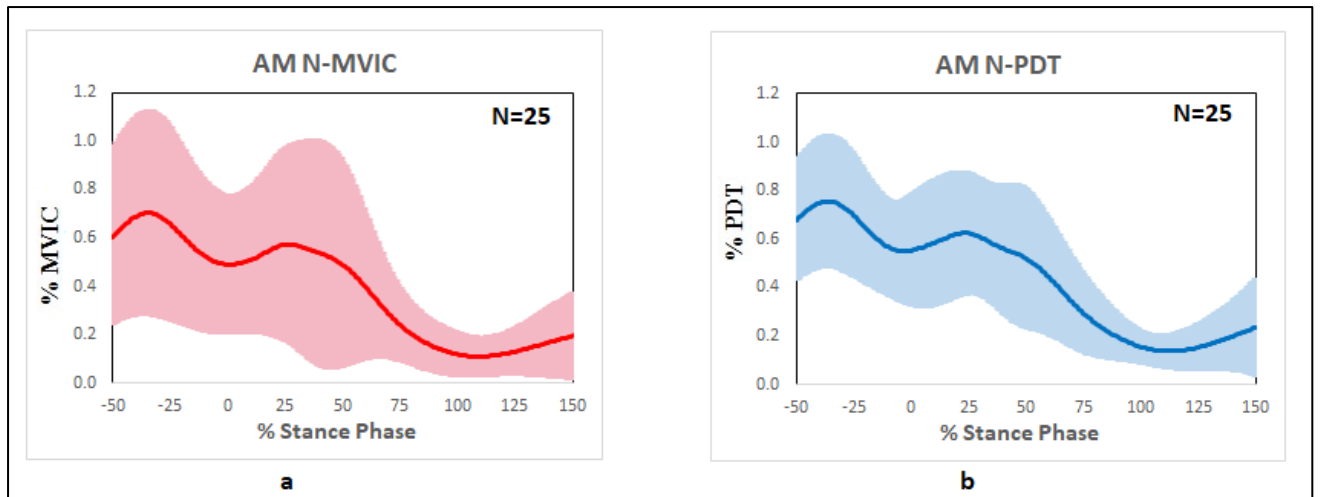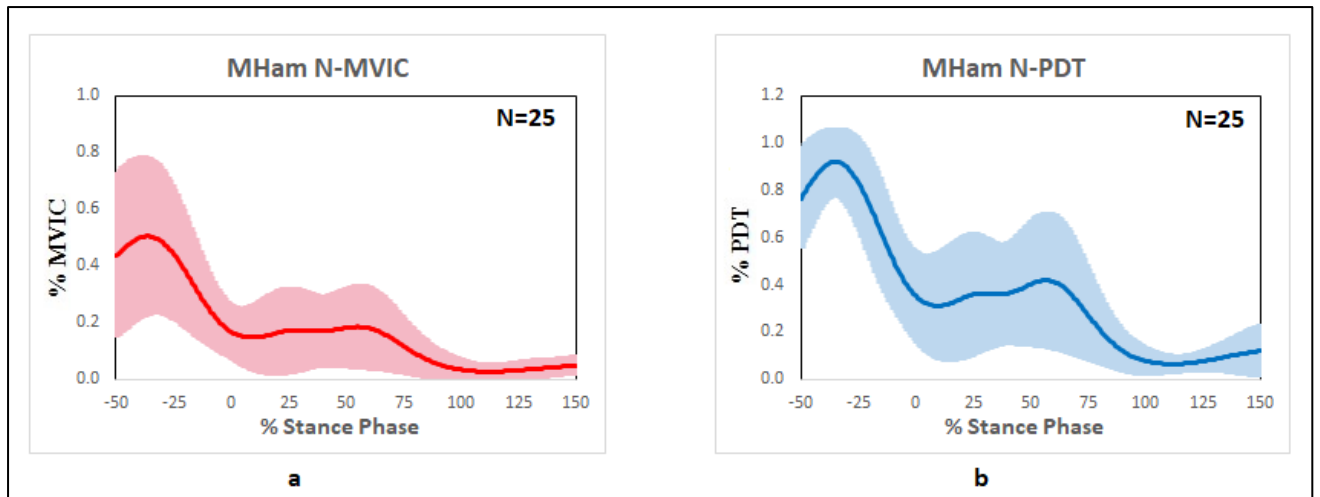
Figure 5- 8: The figure displays the ensemble average (± 1 SD) of: a. medial hamstrings (MHam) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. MHam when normalised to the peak activity of the dynamic trials (N-PDT). 25 participants were used to create the profile of this muscle.
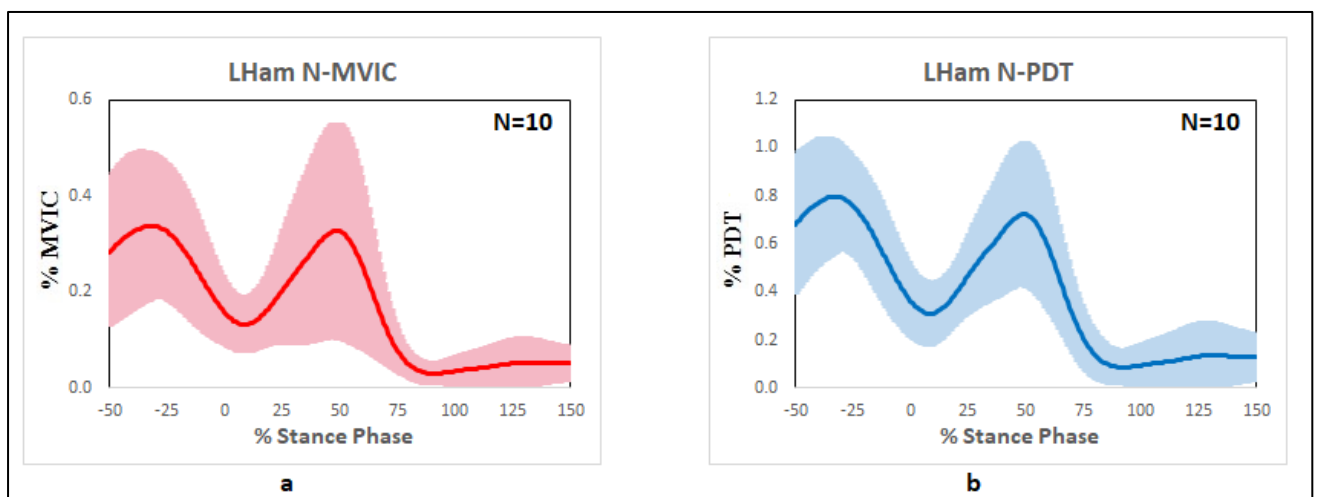
Figure 5- 9: The figure displays the ensemble average (± 1 SD) of: a. lateral hamstrings (LHam) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. LHam when normalised to the peak activity of the dynamic trials (N-PDT). Ten participants were used to create the profile of this muscle.

5.5.2.4 Gastrocnemius group

The EMG profile for MGastro and LGastro started just before the foot contact (about -10% pre-stance) and ended at about 75% of the stance phase. The gastrocnemius group showed a profile with one peak at around 40% of the stance phase for both normalisation methods (Figure 5- 10 and Figure 5- 11). When the amplitudes of MGastro and LGastro were normalised to MVIC, the peaks reached 100% and 80% of MVIC, respectively. In contrast,

the peaks of MGastro and LGastro reached 95% and 85% of PDT when normalised to PDT.

For both gastrocnemius muscles, ensembles normalised to PDT provided the lowest inter-

subject variability compared to ensembles normalised to MVIC. Lower inter-subject

variability was seen at 50% of the stance for both normalisation methods.
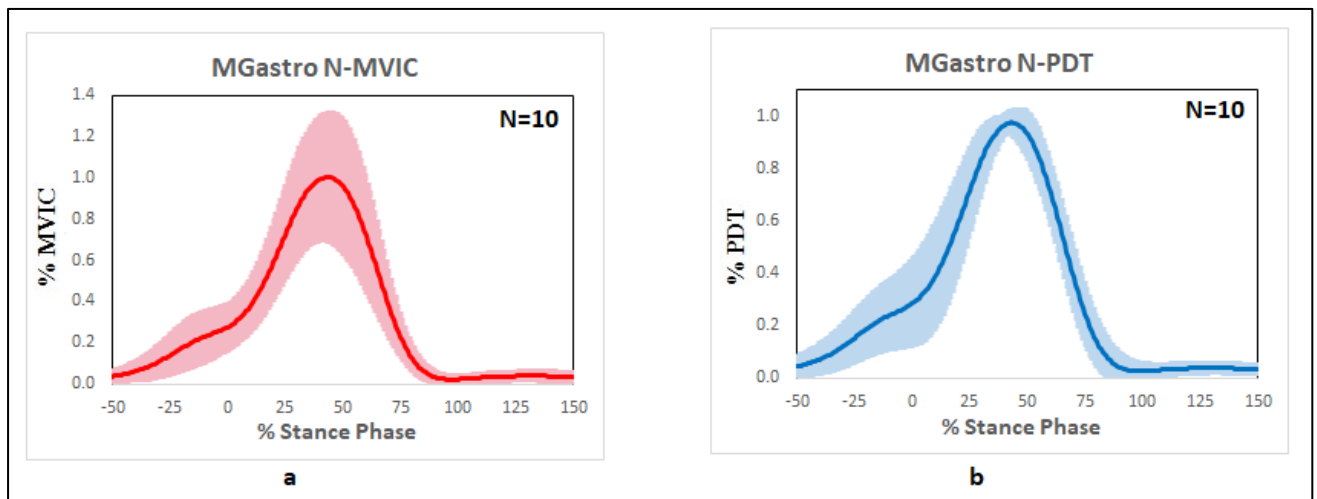
Figure 5- 10: The figure displays the ensemble average (± 1 SD) of: a. medial gastrocnemius (MGastro) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. MGastro when normalised to the peak activity of the dynamic trials (N-PDT). Ten participants were used to create the profile of this muscle.
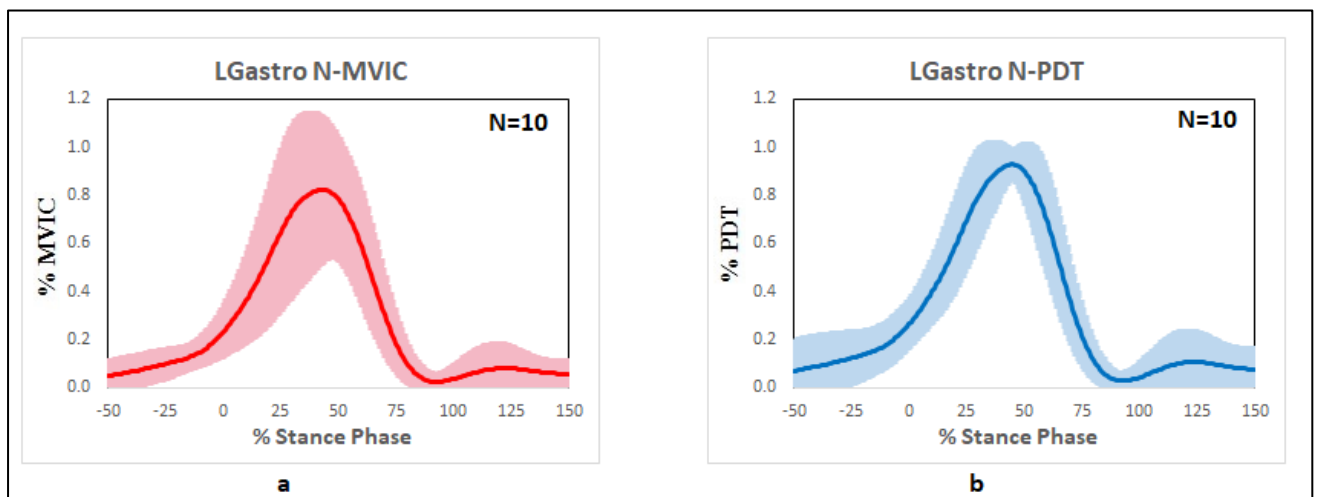
Figure 5- 11: The figure displays the ensemble average (± 1 SD) of: a. lateral gastrocnemius (LGastro) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. LGastro when normalised to the peak activity of the dynamic trials (N-PDT). Ten participants were used to create the profile of this muscle.

5.5.2.5 Tibialis anterior

The activity of the TA extended over the complete stance, starting before foot contact (-20% of the stance) and ending at about 75% of the stance phase, with two peaks. The highest peak occurred at the pre-activation period (about -10% pre-stance) and the other occurred at around 50% of the stance for both normalisation methods (Figure 5- 12). When EMG activity of TA was normalised to MVIC, the peaks reached about 50% and 20% of MVIC. In contrast, the peaks of TA reached 90% and 30% of PDT when normalised to PDT. Ensemble normalised to the PDT provided the lowest inter-subject variability compared to the ensemble normalised to MVIC. For both normalisation techniques, lower inter-subject variability was seen at 0% of the stance.
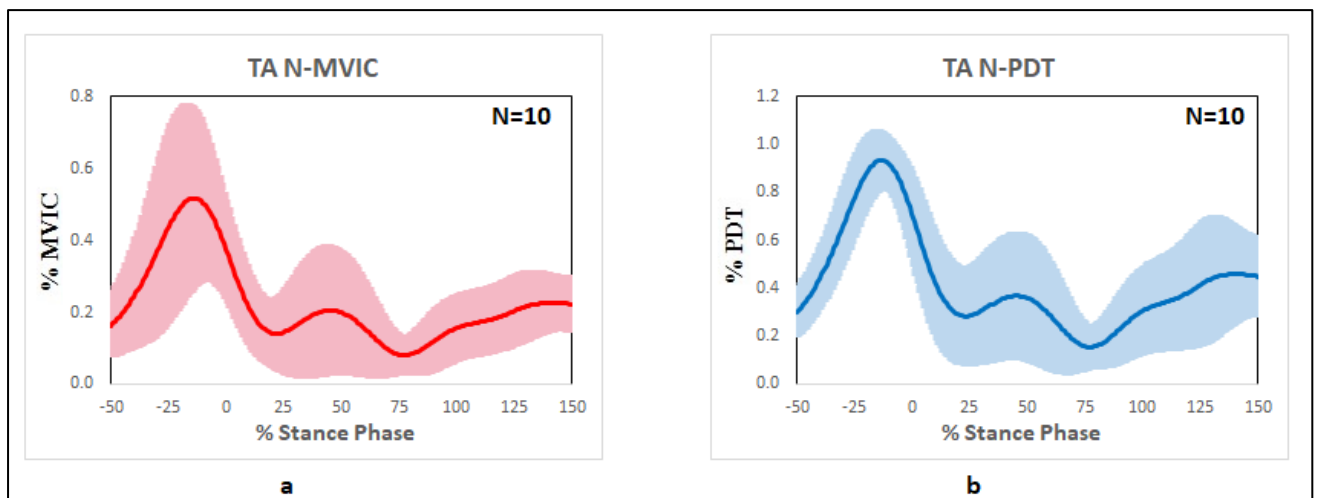
Figure 5- 12: The figure displays the ensemble average (± 1 SD) of: a. tibialis anterior (TA) when normalised to the maximum voluntary isometric contraction (N-MVIC); b. TA when normalised to the peak activity of the dynamic trials (N-PDT). Ten participants were used to create the profile of this muscle.

5.6 Discussion
Key observations

The three adductor muscles displayed different patterns of activity during running. AL peaked around toe-off, while Gr and AM peaked around the mid-stance and the initial contact, respectively (Figure 5- 1, Figure 5- 2, and Figure 5- 3). However, although a general

pattern was present throughout the stance, there was a relatively high level of inter-subject variability for the three adductors. This may be due to the variations in stance phase time, contact time, which could result in variation in the duration of EMG activity for a given muscle. However, the variability in stance time was relatively low (~11%) so would not have a major impact on the activation profiles across the different subjects. Furthermore, variations in amplitude of muscle activation would not explain differences in the level of activation of the different muscles across the cohort studied. Moreover, the adductor profiles indicated that the MVIC normalisation method resulted in higher inter-individual variability compared to the PDT normalisation method. The lowest inter- individual variability for AM occurred at the initial contact phase (0% of the stance) for the amplitudes normalised to MVIC, while for amplitudes normalised to PDT, the lowest variability occurred late in stance (100% of the stance). In contrast, the lowest inter-individual variability for the AL occurred late in the stance (100% of the stance) for the amplitudes normalised to MVIC, while for amplitudes normalised to PDT, the lowest variability occurred at the initial contact phase (0% of the stance). For Gr, the lowest inter- individual variability occurred late in the stance for both normalisation methods.

The EMG activity for GMax, GMed, VMO, and VLO showed very similar profiles, starting before foot contact and ending at about 55-60% of the stance phase, with one peak at 15-20% of the stance phase (Figure 5- 4, Figure 5- 5, Figure 5- 6, and Figure 5- 7). Moreover, the hamstring group activated approximately at the pre-stance period and continued approximately up to 75% of the stance, with two peaks at the pre-stance and at 50% of the stance (Figure 5- 8 and Figure 5- 9). The EMG profile for MGastro and LGastro started around the foot contact and ended at about 75% of stance phase, with one peak at around 40% of the stance phase (Figure 5- 10 and Figure 5- 11). Moreover, the activity of TA extended over the complete stance, with two peaks at pre-activation and at 50% of the stance

(Figure 5- 12). Compared to the PDT normalisation technique, the MVIC normalisation technique produced higher inter-subject variability for all tested muscles. The lowest inter-subject variability occurred at the initial contact phase (0% of the stance) for both normalisation techniques for all muscles apart from the MGastro and LGastro muscles

5.6.1 Adductor group

The results of the current study showed that the three adductor muscles demonstrated different activation patterns during running. AL peaked around toe-off, while Gr and AM peaked around the mid-stance and the initial contact, respectively (Figure 5- 1, Figure 5- 2, and Figure 5- 3). These profiles can be explained by considering the anatomical function of the adductor muscles in the different plane of motion. The adductors muscles exhibited relatively large moment arms for frontal plane moments and also exhibited moment arms of varying magnitudes in the sagittal plane. For example, the AM muscle is actually a very important hip extensor in the early stance phase. This is because AM has a longer moment arm for hip extension compared the main hip extensor muscle (Nemeth & Ohlsen, 1985). Similarly, AL acts to produce hip flexion during the late stance (Dostal et al., 1986). In addition, Gr is a two-joint muscle acting simultaneously on the knee and the hip. Therefore, the inter-subject variation in the activity of adductor muscles could well be a result of synergistic differences (inter-subject difference in coordination) in the activation of these muscles which may result from their role in assisting with sagittal motion and controlling frontal plane movements.

The inter-subject variability of the adductor is large when normalised to MVIC compared to the adductor normalised to PDT. Different inter-subject variability was observed across the adductor muscles either normalised to MVIC or PDT. When the EMG amplitudes were normalised to MVIC, lower inter-subject variability was observed at 0% for AM and at 100%

for AL and Gr. However, when the EMG amplitudes were normalised to PDT, the lower inter-subject variability was observed at 0% for AL and at 100% for AM and Gr. This variability is not necessarily the result of measurement error (discussed earlier in Chapter 4) but it could be a consequence of synergistic variation in coordination patterns between different individuals (as explained above). This may be due to the multiple roles of adductors, which have the capacity to generate force both in the frontal and sagittal planes and the possibility that this multiple function alters coordination balance between adductors and muscles in other planes.

In the current study, AM demonstrated a profile starting from the pre-activation period and ending at about 75% of the stance, regardless the method of normalisation. This is consistent with Perry (2010) who examined AM muscle activity. They showed that AM activity was increased at the initial contact to absorb some of the shock of ground contact while preserving progression and to provide postural stability by optimally positioning the limb. Additionally the results of the current study showed the AM profile with a peak at early stance, which coincide with Montgomery et al. (1994) who proposed that AM peak occurs early in the stance during the loading response. Similarly, Wall-Scheffler et al. (2010) reported that the hip adductors are active throughout the stride, and the activity increases across the gait cycle. As AM plays different roles in different planes (sagittal and frontal), its profile demonstrates higher inter-subject variability.

The results of the current study showed that the AL activated later in the stance phase (at about 25%) and continued throughout the stance phase, with one peak around the toe off. In contrast, Mann et al. (1986) found that the activation period for AL began just after toe off and continued into early forward swing. This early activation could be due to the higher speed used (4.5 m.s^{-1}), with a different method for recording the muscle activity (indwelling

electrodes) in the previous study. In contrast to the finding of the current study, Chaudhari et al. (2014) reported that the activity of the AL started at the pre-contact phase and continued to the toe off. This difference could be due to improper electrode placement in the previous study, thus increasing the possibility of crosstalk from the adjacent muscles. The earlier activation of AL at early and mid-stance could occur as a result of an external hip abduction moment (Chaudhari et al., 2014).

5.6.2 EMG profiles of the major lower limb muscles during the stance phase of running

5.6.2.1 Gluteal group

The results of the current study showed that the GMax and GMed profiles started before foot contact to midstance, which was similar to AM activity. This similarity in the profiles may be attributed to the different functions of AM exhibited by its different portions. For example, during unilateral weight-bearing, the heavy trunk is liable to move in any direction on top of the ball-and-socket joint at the hip. Especially, relevant is when the trunk is forwardly inclined, as the hip becomes markedly flexed and the longer portions of the AM (middle and bottom portions) are positioned to strongly rotate the pelvis posteriorly (Takizawa et al., 2014). However, as forward acceleration in running is produced by hip extensors, there are some doubts that the strongest hip extensor (GMax) has a predominant role in this extension. Additionally, GMax has an abducting effect on the leg. These extension and abduction effects of GMax and GMed on the leg could have a negative effect on the straight movement of the support leg from front to back. However, outward rotation and abduction would not be of any significance during running if another muscle could act together with GMax, both to support GMax during the hip extension and also to neutralise the abducting effect of GMax and GMed. This task could be taken over by AM (Wiemann & Tidow, 1995). This was proven by the lower inter-subject variability, observed in the current study, at initial contact in both

GMax and GMed muscles, revealing the presence of other muscles assisting the function of the two muscles (GMax and GMed). This role was played by the AM activity, which activated at the same period of the stance phase.

Our findings are consistent with the findings of Novacheck (1998) and Gazendam and Hof (2007). They suggest that the GMax activity functions to prepare the limb for ground contact and to absorb the shock of that impact during the stance phase of absorption. It is proposed that the GMax activity helps to extend the hip during this period (Kyröläinen, Avela, & Komi, 2005; Kyröläinen, Belli, & Komi, 2001; Novacheck, 1998). At initial foot contact, GMax is contracting eccentrically to limit hip flexion and to stabilise the stance limb. All the way through to take-off, GMax is concentrically moving the limb into extension (Beard, 2015). The maximum hip extension mostly occurs at or just after toe off (Schache, Bennell, Blanch, & Wrigley, 1999), which is primarily facilitated by GMax (Beard, 2015). Therefore, the GMax contributes to hip extension during early stance and propulsion function at late stance (Ellis, Sumner, & Kram, 2014).

Similar to the GMax, GMed activated just before the initial contact and ended at about 55% of the stance phase. This finding is consistent with the work of Novacheck (1998) and Gazendam and Hof (2007). The activity of GMed occurs to control the hip adduction during the stance phase and to provide supplemental propulsion, supporting experimental studies (Bartlett, Sumner, Ellis, & Kram, 2014; Ellis et al., 2014; Novacheck, 1998). As we mentioned before, GMax has an abducting effect on the leg which may explain the same profile activity during the stance phase. During the stance phase of running, GMed is responsible for maintaining a neutral and stable pelvis (Dicharry, 2010). At the initial contact phase, the GMed attempts to control the degree of hip adduction generated by the hip adductors. As the limb moves into the mid-support phase, the GMed is acting eccentrically to

maintain a level pelvis from which the swing leg moves. At take-off, the GMed contracts concentrically to create hip abduction (Beard, 2015). Poor frontal plane pelvic control may contribute to atypical activity in GMed or hip adductors. This will increase the hip adduction excursion while running and contribute to several running-related injuries (Semciw, Neate, & Pizzari, 2016).

5.6.2.2 Quadriceps group

The finding of the current study demonstrated that the EMG activity for VMO and VLO started before foot contact and ended at about midstance to provide primary propulsion. This is very similar to what has been suggested by Gazendam and Hof (2007) Novacheck (1998), and Ellis et al. (2014). The activity of the quadriceps occurs as a preparation for ground contact. It acts as braking force providing the primary means of shock absorption. The quadriceps muscle then works eccentrically to resist knee flexion from mid-support to take-off (Beard, 2015). The greatest activity of VMO and VLO occurred between the pre-contact and braking phases. In these phases, the activity of both vasti could exceed MVIC, while in the propulsive phase, their EMG activity could reach below 50% of MVIC (Kyröläinen et al., 2005). The activity of both vasti, however, almost disappeared early before the toe off (Kyröläinen et al., 2001), and only the RF continued the activity in midswing (Novacheck, 1998). This is essential to restrain the posterior movement of the tibia as the knee flexes.

The starting point of the stance phase (0%) generally had the least inter-subject variability for both normalisation techniques for the quadriceps. These values, compared to data reported by Guidetti, Rivellini, and Figura (1996), were lower than values referring to running EMG data either for the normalised mean ensemble value or even the peak ensemble value for VMO and VLO. The lower inter-subject variability could indicate better and more accurate

electrode placement used in the current study (SENIAM guidelines). Unlike the previous study, the EMG electrodes were simply placed over the muscle belly.

5.6.2.3 Hamstring group

The results of the current study generally agree with previous findings on the EMG profiles for the hamstring group, which showed that MHam and LHam activated pre-stance and continued up to 75% of the stance phase, with two peaks occurring during the swing phase and at around 50% of the stance phase (Gazendam & Hof, 2007; Novacheck, 1998). During part of this time, the knee is flexing, the hip is extending, and the hamstrings (two joint muscles) may be acting to extend the hip and control the knee. The dual role of hamstrings on hip and knee could result in higher inter-subject variability of the motor control pattern for this group of muscles. Peak MHam and LHam activity occurred at the late swing phase, with minimal stance phase activation supporting the idea that their primary function during running is arresting the swing leg (Ellis et al., 2014). In addition, the activity of the hamstring muscle prepares the stance limb for ground contact by decelerating the rapidly extending knee (Ellis et al., 2014; Novacheck, 1998). The highest peak occurring at the pre-stance could reach a level equal to the maximal voluntary contractions (Kyröläinen et al., 2005). Lowest variability between subjects was displayed at the initial contact in the current study. Compared to the current study, Guidetti et al. (1996) reported higher inter-subject variability at the same period of the stance. The reason for lower variability observed in the current study could be due to better electrode placement. They simply placed the surface EMG electrodes over the muscle belly, while the SENIAM guideline was used in the current study.

5.6.2.4 Gastrocnemius group

The gastrocnemius behaves in a similar way to GMax and the quadriceps muscles. The observations in the current study revealed that MGastro and LGastro activated just before the foot contact and ended at about 75% of the stance phase. Such profiles support the idea that the gastrocnemius functions for propulsion during running (Ellis et al., 2014). In addition, the results of the current study showed that the gastrocnemius activity had a single peak similar to the quadriceps peak (at around the 40% of the stance phase). This is similar to what was reported by Kyröläinen et al. (2005); Novacheck (1998), and Gazendam and Hof (2007), who found that the profile of the gastrocnemius started shortly before stance and ended before toe off. During initial contact, LGastro and MGastro are eccentrically contracted to help in absorption of the impact. After initial contact and through mid-support, the centre of mass falls medial to the stance limb, forcing the gastrocnemius and soleus to work eccentrically to stabilise the subtalar joint and limit excessive pronation. From mid-support to take-off, the gastrocnemius is the primary generator of the anterior propulsive energy (Beard, 2015).

For both MGastro and LGastro, lower variability was seen at 50% of the stance. This variability is attributed to the differences in the foot strike pattern at initial foot contact. The contact pattern results in different amplitudes and timing of gastrocnemius activation profile. This is supported by Shih et al. (2013), who found that the EMG amplitudes of the gastrocnemius increased at initial contact compared to its pre-activation level in RFS runners (about four times), while FFS runners showed a slight increase in the activation pattern (about 28%). Moreover, Ahn et al. (2014) found that FFS runners activated their plantar flexor muscles 11% earlier and 10% longer than RFS runners at the beginning of the stance.

5.6.2.5 Tibialis anterior

The TA profile started before foot contact (-20% of stance) and ended at about 75% with two peaks: the highest peak occurred in the pre-activation period (at about-10% pre-stance) and the other occurred at around the 50% of the stance. This result was found to be in accordance with Novacheck (1998), who found that the activity of TA during running extended from the pre-contact and ended at about 75% of the stance phase. In the late swing, TA concentrically dorsiflexes the ankle in order to allow ground contact with the hindfoot initially and then, eccentrically control the lowering of the forefoot to the ground during the first part of the stance (Novacheck, 1998). However, such control of the foot slap is absent in a forefoot striker (Loudon, Manske, & Reiman, 2013). Moreover, the results of the current study agreed with the findings of Gazendam and Hof (2007) who reported that the prominent peak was at 90% and its minor activity was present in the first half of the stance. Lower inter-subject variability for TA activity was observed at the initial contact (0% of stance). Similar to the gastrocnemius, the source of variability in the TA activation pattern could be attributed to the foot strike pattern.

5.7 Limitations

One of the limitations associated with this study is that it was limited to young fit lean runners and this cohort may not capture the true variability across the demographic spectrum. Nevertheless, a high level of variability was seen in the adductor muscles. This could be due to the nature of these muscles and the role that they could play in both the sagittal and frontal planes of movement. Moreover, it was not possible in the current study to measure the activity of the hip flexor due to its deep position and the need to use another measuring tool (fine wire) which was not available. Finally, the protocol of this study was limited to one

speed as investigating the effect of running at different speeds on the muscle activation pattern was beyond the scope of this thesis.

5.8 Conclusion

The current study aimed to describe the typical EMG profile for the adductor muscles during the stance phase of overground running. In addition, it was aimed to describe the typical EMG profile for the major lower limb muscles collected during running and their associated inter-subject variability. The most important finding in this study is that the activation pattern of lower limb muscles works with certain synchronised mechanisms, and any source could break this mechanism will lead to atypical activation motor control of these muscles. Muscles such as AM, AL, GMax, and GMed that have a role in multiple planes could have high inter-subject variability in their profiles. Similarly, the two-joint muscles such as Gr, MHam, LHam, MGastro, and LGastro could have a higher level of between-subject variability during running.

Chapter 6: Association of the frontal plane pelvic motion and the adductor activation pattern during running

6.1 Introduction

The previous chapters of this thesis describe the development of a robust protocol for quantifying adductor muscles activity during walking and running using surface EMG electrodes. This protocol demonstrated good to excellent between-day reliability in both walking and running (for more details see chapter 4). Using this protocol, AM and Gr were observed to activate at foot strike with one peak occurring early in stance during running. In contrast, AL showed a different profile during running. It activated late in stance with one peak around the toe off. Interestingly, although there was a characteristic pattern of activation for each of the three adductor muscles, there was substantial inter-subject variation. It is possible that this variability could be associated with kinematic differences in running style. In line with this idea, in this chapter, the link between the pelvic drop (one of the most common risk factors linked to running related injuries) and the pattern of activation of the adductor muscles will be explored.

Pelvic drop refers to the kinematic pattern, in which the pelvis drops (in the frontal plane) away from the stance limb. Technically, this is referred to as a downward obliquity, or Trendelenburg sign, of the opposite hip relative to horizontal during its swing phase (Figure 6- 1).
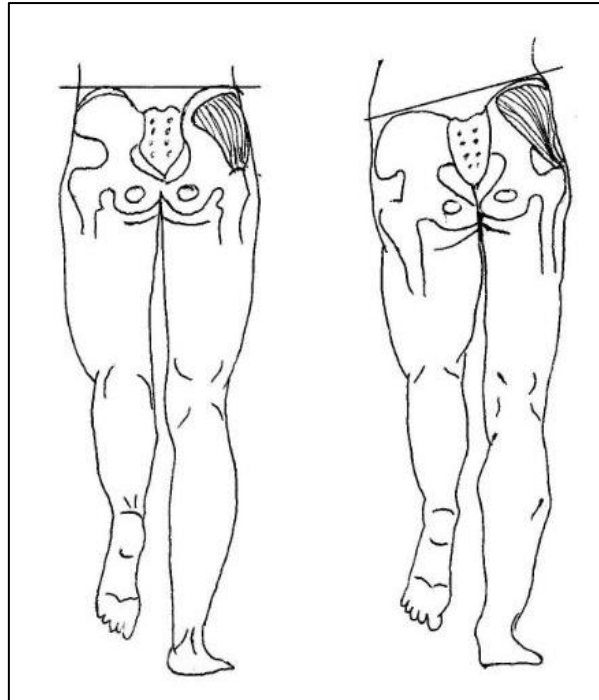
Figure 6- 1: Example of right frontal plane pelvic drop. The runner is in right stance phase, and the pelvis is rotating in the frontal plane about the right hip, such that the left PSIS has dropped below horizontal.

As previously discussed in the literature review, CPD and excessive hip adduction have been suggested as risk factors for many RRI (Fredericson et al., 2000; Noehren et al., 2007; Pohl et al., 2008). Recreational runners with patellofemoral pain syndrome, tibial stress fracture, and iliotibial band syndrome, tend to display greater hip adduction relative to healthy controls (Fredericson et al., 2000; Noehren et al., 2007; Pohl et al., 2008). Significantly, CPD has also been shown to be a risk factor for lower limb pathologies that are non-running related, such as osteoarthritis (OA). Park et al. (2010) and Hinman et al. (2010) observed the presence of excessive hip adduction angles in knee OA patients. Moreover, Allison et al. (2016) found that individuals with gluteal tendinopathy exhibited greater hip adduction moment, which was associated with pelvic drop and contralateral trunk lean during walking. This highlights the importance of the CPD as an important kinematic pattern that needs to be studied in greater depth (Bramah, Preece, Gill, & Herrington, 2018).

It has been proposed that pelvic drop is the result of weak hip abductor muscles (Fredericson et al., 2000). Theoretically, running is mostly a sagittal plane activity; therefore, muscles associated with the frontal plane could become weakened without cross-training or strengthening (Burnet et al., 2008). In support of this idea, Fredericson et al. (2000) measured hip kinematics following a six-week hip-strength programme with a focus on GMed strengthening. They concluded that the increased strength of hip abductors was associated with a reduction in hip adduction angle in distance runners with iliotibial band syndrome. Interestingly, similar results were obtained for non-running related injuries, such as knee OA (Park et al., 2010). Previous studies have suggested that strengthening the GMed fosters increased control of thigh adduction tendencies, thereby minimising the valgus vector at the knee. However, other research does not corroborate this idea; for example, Willy and Davis (2011) and Baggaley et al. (2015) found no correlation between hip abduction strength and either peak hip adduction or pelvic drop during running. Moreover, they determined that there was no relationship between GMed isometric torque and frontal plane pelvic drop in recreational runners. Importantly, it is unclear whether muscles measured as weak in static tests will necessarily exert less force during dynamic activities, such as running.

Both adductor and abductor muscles generate frontal plane hip movements. Moreover, the way the pelvis moves in the frontal plane is determined by a range of biomechanical factors, including the relative positioning of the foot, centre of mass and hip joint centre, as well as the activities of a range of muscle groups including the adductors and abductors. However, to date, previous research has focused on the abductor group, with minimal focus on the adductor group. Importantly, an over-activation of the adductor muscles could increase the adduction movement at the hip and therefore increase CPD. However, this has not yet been investigated. Therefore, further research is required to understand the potential association between adductor activity and pelvic motion in the frontal plane. It is possible that the

insights gained from such a study will identify pathomechanics, which in turn may play a critical role in injury prevention and management. As discussed earlier, most researchers and therapists focus their attention on abductor muscle activity and strength as the main causes of the pelvic drop in various types of pathologies. Interestingly, there has been no study which has investigated the activity of the adductor muscles during the stance phase of running, and how this activity could affect frontal plane pelvic movement. Therefore, the current study aims to investigate the association between the frontal plane movement and muscle strength, hip moments and the EMG activity of the lower limb muscles at a specific point of the stance phase of running.

6.2 Aim

This study aimed to investigate the association between the frontal plane pelvic movement and the muscle strength, hip moments and the EMG activity of the lower limb muscles during the early stance phase of running.

6.3 Methodology

This section demonstrates the current study procedures. This includes the recruitment plan, inclusion and exclusion criteria, the instrumentation used and the testing protocol.

6.3.1 Recruitment plan

Similar to the recruitment plan applied in the chapter 4, the participants were invited to participate in this study through a number of avenues (see chapter 4; section 4.3.1). This included placing posters around the university campus, emailing the local running/ triathlon clubs and advertising via the running performance clinic website. A number of responses from many runners were received and only those who met the entry criteria were purported for participation in this study.

6.3.2 The entry criteria

The same inclusion criteria which applied in the previous chapter were applied in this study as well (see chapter 4; section 4.3.2). All participants were males aged between 18-40 years. The participant had to be free from lower limb injuries in the past six months. In addition, the training routines for all participants had to include running training at least three times per week for a minimum of 10-15 miles per week. This training programme was performed for at least three months prior to enrolment in the study.

6.3.3 Participants

The same participants tested in the previous study were used in this study for further analysis. This included a cohort of 25 male runners, with no history of lower limb injury or surgery. The participants characteristics were presented in chapter 5, section 5.3.3.

6.3.4 Ethical approval

Before starting the data collection, all participants read and signed a written informed consent statement approved by the Research, Innovation and Academic Engagement Ethical Approval Panel at the University of Salford (Appendix IV).

6.3.5 Instrumentation

Similar to the previous studies, an ultrasound imaging system (MyLab70, Esaote, USA), a Direct Transmission System with 16 channels (Noraxon USA inc., model 586 Tele Myo DTS Desk Receiver), the DTS sensors (model 542), EMG lead (542AP) and a disposable adhesive Ag/AgCl EMG electrode (for more details, see the methodology section in the chapter 3) were used. In addition, a motion capture system (ten Pro-Reflex, Qualisys cameras with three embedded force platforms) was used in this study in order to identify the gait events (for more details, see the methodology section in chapter 4). Also, all set up procedures for the

ultrasound, EMG measurements, and motion capture system followed the steps undertaken in the previous studies (see chapter 3; sections 3.4.3 and 3.4.5, and chapter4; section 4.3.5).

6.3.6 Surface EMG electrode and 3D marker placement

After signing the consent form, the participants were asked to change into their shorts and a comfortable t-shirt. Then, the height and weight of the participants were measured. This was followed by *placement of the surface EMG electrodes*. Similar to chapter 4, the muscles were located according to the SENIAM guidelines (Hermens et al., 1999). The EMG amplitudes were obtained from seven muscles on each side of the body: GMax, GMed, VMO, AL, Gr, AM, and MHam. The adductor group (AL, Gr, and AM) was located similar to the approaching techniques used in study 1 (see chapter 3; table 3-2). All undertaken steps for skin preparation and the signal checking process followed in chapter 3 were followed in this study (see chapter 3; section 3.4.5).

Marker placement. Forty reflective markers of 14.5 mm diameter were used in all trials of data collection. The markers were attached to the skin using hypoallergenic double-adhesive tape attached to a flat-based marker. The marker set was distributed over the different lower limb parts similar to what was applied in the previous study (see chapter 4; section 4.3.6).

6.3.7 Testing protocol

The procedures of the testing protocol undertaken in the previous chapter were followed in this chapter as well (see chapter 4; section 4.3.7). This included the protocol for recording the static and dynamic trials, as well as the protocol for obtaining the MVIC. The adductor muscles were normalised to the MVIC according to the results of the pilot study undertaken in chapter 4 (see Chapter 4; Table 4.1).

A handheld dynamometer was used to measure the strength of the abductor and adductor muscles. With this protocol, both right and left sides were assessed using the dynamometer with the limb in the same positions used for measuring the EMG amplitudes during MVIC testing (for more details see section 4.3.7.3). The participant was asked to gradually build up his maximum contraction against the manual resistance of the examiner who held the handheld dynamometer against the distal end of the femur. The participants were then instructed to hold a maximum contraction for a minimum of five seconds. This process was repeated 3 times for each tested position with a 30 seconds rest in between. Maximum peak force was recorded throughout the three trials. The handheld dynamometer demonstrated good intrarater and interrater reliability for isometric hip and knee muscles strength (Arnold et al. 2010).

6.4 Data processing
6.4.1 Three-dimensional processing

Similar to the processing plan for the 3D measures which was undertaken in chapter 4, the kinematics and kinetic variables were processed in this chapter (see chapter 4; section 4.4.2). This included markers labelling and filling any trajectory gaps with maximum consecutive 10 frames in the QTM software, filtering the motion and force plate data and calculating the joint kinematic and kinetic data using V3D software.

The main outcomes for the current study were the frontal plane pelvic movement, the sagittal and frontal hip moments, hip strength and EMG activity for the muscles surrounding the pelvis. Initially, the peak pelvic drop angle for each side was measured individually for each participant during the stance phase. Then, the other outcomes were matched to this peak

angle. For the strength measure, each participant performed three testing trials for each testing position and the three trials were then averaged for each position.

6.4.2 EMG and MVIC processing.

The processing plan of the EMG data which was applied in the previous study was followed in this study (see chapter 4; section 4.4.1). This included time normalisation, removal of movement artefacts, and averaging the gait data. In addition, the *MVIC* was processed similarly to that performed in the previous study (see chapter 4; section 4.4.1). In order to synchronise between the EMG activity and lower extremity motion, there is a need to specify the electromechanical delay (EMD). This period is defined as the time between the onset of the myoelectric signal and the initiation of muscle tension. This interval is assumed to represent the propagation of the action potential along the muscle, the excitation-contraction coupling process, and stretching of the muscle's series elastic component by the contracting component (Zhou, Lawson, Morrison, & Fairweather, 1995). This delay is important to consider if the researcher wants to precisely relate EMG and motion in research studies.

A number of different studies have proposed estimates for the EMD which could be used to interpret gait data. However, each study appears produces a slightly different estimate. For example, voluntary knee extension has been found to produce an estimate of EMD=40 ms (Perry, 1998). Similar estimates have been reported by Houston, Norman, and Froese (1988) who observed an EMD of 43.3 ms and Zhou et al. (1995) who observed an EMD of 38.7 ms for knee extensors. Based on these findings, Perry (1998) suggested that the average EMD during gait is no more than 40 ms. Therefore, in the current study, the EMD was set as 40 ms. This value of 40ms was used to adjust the time period of interest backwards in order to factor in EMD.

The precise window of EMG was determined individually for each separate participant and. calculated as follows. Firstly, the timing of the peak pelvic drop angle was identified for the individual participant (for both the left and right sides) and adjusted backwards to account for EMD (see above). The MVIC normalised EMG data for each muscle was then averaged across this window to produce muscle-specific outcomes used for the statistical analysis, see below. This process was repeated separately for each participant.

6.5 Statistical analysis

Firstly, the assumptions of parametric statistical tests were checked through the Shapiro-Wilk test for normality for all variables. All data were determined to meet the assumptions for parametric testing. In order to investigate the association between the pelvic drop and the selected biomechanical variables during the early stance phase of running, bivariate correlations were studied by using the Pearson Correlation Coefficient (r) with a significant level of 0.05). In order to determine the strength of the association, the following guide was used: r=0-0.19 was regarded as very weak, r=0.2-0.39 as weak, r=0.40-0.59 as moderate, r=0.6-0.79 as strong and 0.8-1 as very strong correlation (Campbell & Swinscow, 2011). All statistical analysis was carried out using SPSS for Windows version 24.0 (SPSS Inc., Chicago, IL) and Excel (Microsoft Office Excel, 2016). Additionally, the pelvic frontal plane movement was divided into right and left CPD. Each side and its associated biomechanical variables were analysed separately.

6.6 Results

The descriptive statistics mean (SD) for the tested variables (hip strength, hip moments and EMG for lower limb muscles) which were included in the correlational analyses for right and left sides are provided in Table 6- 1.

Table 6- 1: The descriptive statistics mean (SD) for the abductor and adductor muscles strength, sagittal and frontal plane moments and EMG for lower limb muscles for the right and left sides.

Variables		Right side		Left side	
		Mean	SD	Mean	SD
Muscle strength (Nm/Kg)	Abductors	1.84	0.29	1.81	0.29
	Adductor 0°	1.05	0.19	1.03	0.19
	Adductor 45°	1.32	0.25	1.33	0.17
Moment (Nm/Kg)	Sagittal plane	0.16	0.59	0.03	0.44
	Frontal plane	1.64	0.54	1.75	0.35
EMG (mV)	AL	0.25	0.26	0.40	0.34
	AM	0.55	0.36	0.45	0.19
	Gr	0.60	0.78	0.67	1.07
	GMed	0.42	0.26	0.36	0.29
	GMax	0.27	0.18	0.26	0.13
	MHAM	0.18	0.12	0.13	0.06

6.6.1 The association of the CPD and the hip strength

Table 6- 2 shows the correlation between the CPD and the hip abductor and adductor strength of both sides. On the right side, the results show that there are non-significant very weak negative correlations between the CPD and the abductor and adductor (at 0° and 45°) muscle strengths. On the left side, the results reveal non-significant very weak negative correlations

between the CPD and the abductor and adductor (at 45°) muscle strengths. However, there is a non-significant very weak positive correlation between the CPD and adductor (at 0°) muscle strength. These data show very little interdependence between muscle strength and CPD.

Table 6- 2: The association between the CPD and the isometric abductor and adductor muscle strength.

Side	Correlation	r	*P* value
Right side	**CPD vs Abductors**	-0.05	0.82
	CPD vs Adductor 0°	-0.02	0.94
	CPD vs Adductor 45°	-0.08	0.71
Left side	**CPD vs Abductors**	-0.13	0.53
	CPD vs Adductor 0°	-0.07	0.73
	CPD vs Adductor 45°	0.04	0.85

6.6.2 The association of the CPD and the hip moments

Table 6- 3 shows the correlation between CPD and the sagittal and frontal plane moments of both sides. On the right side, there is a significant strong positive correlation between CPD and the sagittal plane moment (Figure 6- 2), while the frontal plane moment shows non-significant weak positive correlation. On the left side, the results reveal a significant strong positive correlation between CPD and the sagittal (Figure 6- 3) and a significant moderate positive correlation between CPD and frontal plane moments.

Table 6- 3: The association between CPD and the sagittal and frontal plane moments.

Side	Correlation	r	*P* value
Right side	**CPD vs hip sagittal plane moment**	0.60	**0.001***
	CPD vs hip frontal plane moment	0.23	0.271
Left side	**CPD vs hip sagittal plane moment**	0.63	**0.001***
	CPD vs hip frontal plane moment	0.47	**0.017***
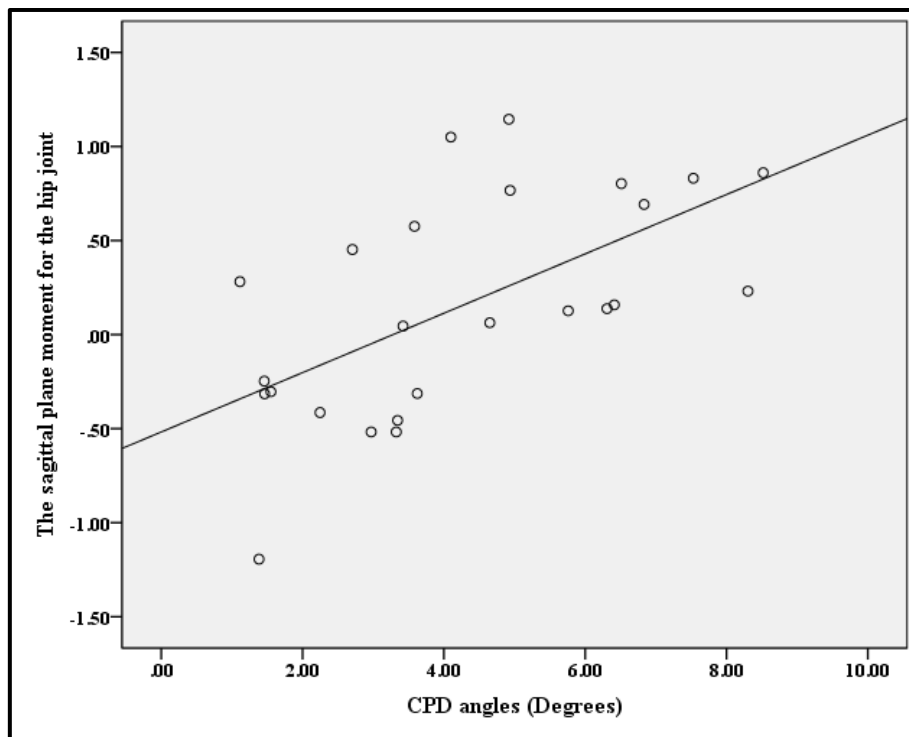
Figure 6- 2: Scatter plots showed the correlation between the contralateral pelvic drop angle (CPD) and the sagittal plane moment for the right hip.
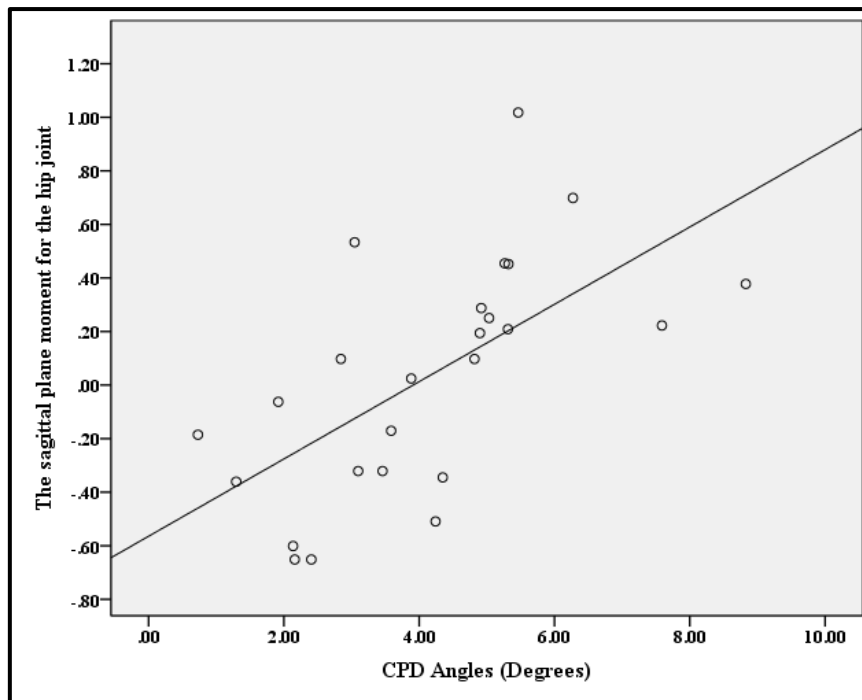
*Figure 6- 3: Scatter plots showed the correlation between the contralateral pelvic drop angle
(CPD) and the sagittal plane moment for the left hip.*

6.6.3 The association of the CPD and EMG for lower limb muscles

Table 6- 4 shows the correlation between CPD and the EMG for lower limb muscles of both
sides. On the right side, a significant strong positive correlation exists between CPD and AM.
However, GMax, MHAM, AL and GMed show a significant moderate negative correlation
with CPD and Gr show non-significant weak negative correlation. On the left side, the results
again reveal a significant strong positive correlation between CPD and AM, while GMax and
MHAM show a significant moderate negative correlation with CPD. The GMed shows a
significant moderate positive correlation, while the Gr and AL show non-significant weak
positive correlation. Figure 6- 4 and Figure 6- 5 displayed the strong positive correlation
exists between CPD and AM for both right and left sides.

212

Table 6- 4: The association between the CPD and EMG for lower limb muscles of both sides.

Side	Correlation	Muscles	r	*P* value
Right side	**CPD vs right side muscles**	**AL**	-0.41	**0.04***
		AM	0.64	**0.00***
		Gr	-0.30	0.15
		GMed	-0.43	**0.03***
		GMax	-0.50	**0.01***
		MHAM	-0.48	**0.02***
Left side	**CPD vs Left side muscles**	**AL**	0.39	0.05
		AM	0.65	**0.00***
		Gr	0.32	0.12
		GMed	0.49	**0.01***
		GMax	-0.41	**0.04***
		MHAM	-0.41	**0.04***
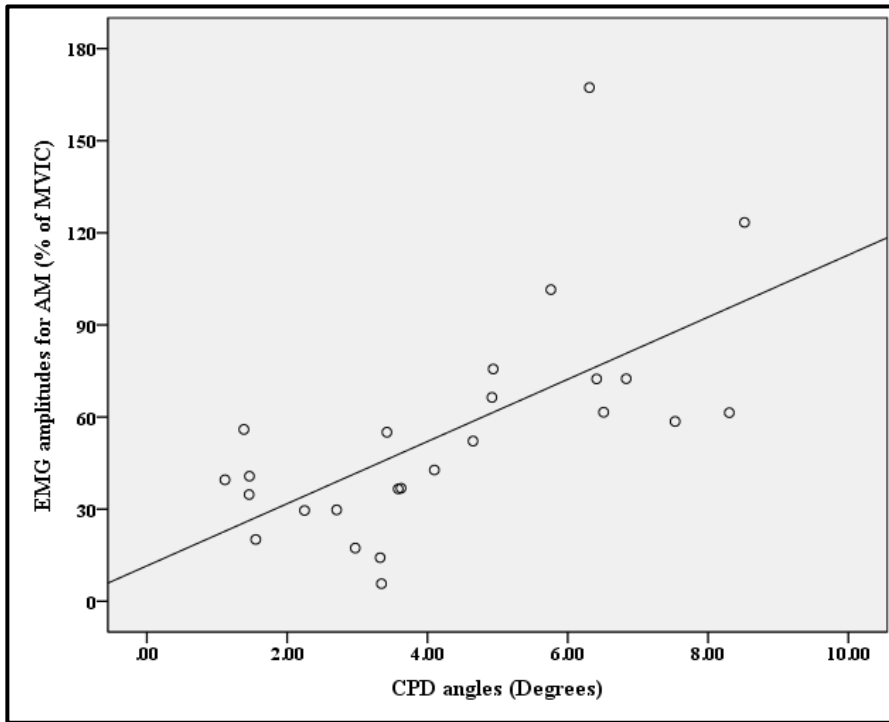
Figure 6- 4: Scatter plots showed the correlation between the contralateral pelvic drop angle (CPD) and the EMG amplitudes of AM when normalised to MVIC for the right side.
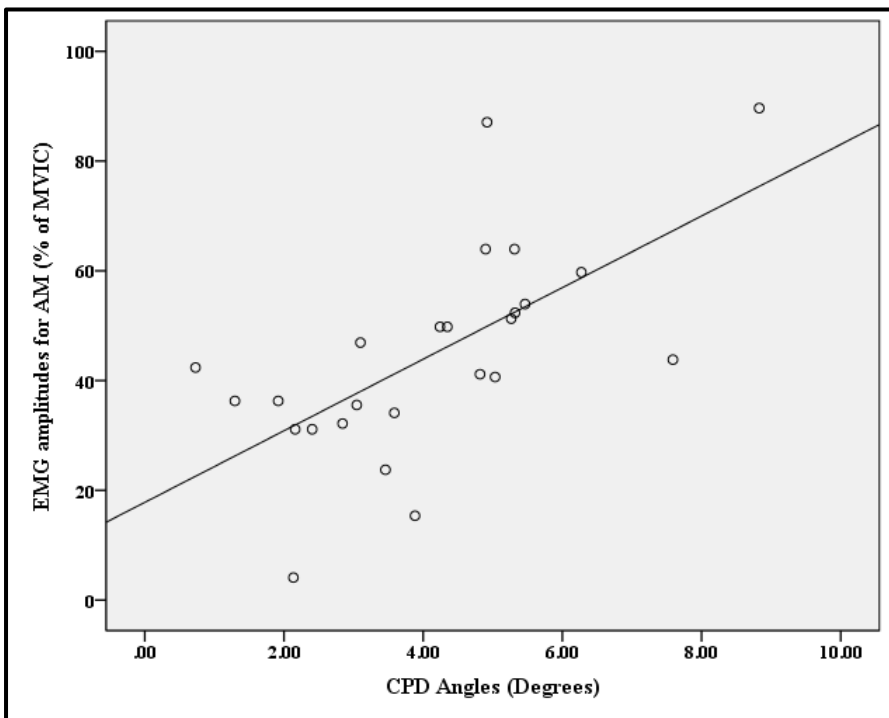
Figure 6- 5: Scatter plots showed the correlation between the contralateral pelvic drop angle (CPD) and the EMG amplitudes of AM when normalised to MVIC for the left side.

214

6.7 Discussion

The primary aim of the current study was to investigate the association between the frontal plane movements of the pelvis and the EMG activity of the lower limb muscles during the early stance phase of running. In addition, the study also sought to investigate possible links between pelvic drop and the hip moments and muscle strength. The most interesting finding of this study was that, on both sides, increased pelvic drop appeared to be strongly associated with increased AM activity. Moreover, there were a number of other differences in muscle activation which, although moderately correlated, did appear to be significantly associated with frontal plane pelvic movements.

6.7.1 The association of CPD and the hip strength

It has been proposed that subjects with decreased isometric GMed muscle strength will exhibit excessive frontal plane pelvic drop (Fredericson et al., 2000). However, the results of the current study show that there were no significant correlations between CPD and either abductor or adductor isometric muscle strength. Thus, the current thinking in the field, which focuses on the idea that weakness in abductor strength is the cause of pelvic drop is not supported by the data in this current study.

In order to investigate whether isometric strength of the abductors could influence frontal plane pelvic movements, previous investigators have sought to understand the effect of abductors strengthening programmes. For example, Park et al. (2010) found hip adduction angle to decrease with increasing the hip abductor strength after a 6-week hip strengthening programme for OA patients. However, these authors did not report pelvic drop angle, nor did they report EMG data for the adductor or the abductor muscles. In another study, Willy and Davis (2011) found that a hip strengthening programme produced non-significant changes in hip mechanics during running. In line with these findings, Burnet et al. (2008) and Burnet and

Pidcoe (2009) showed no relationship between GMed isometric torque and frontal plane pelvic drop across a number of periods during a 30-minute run. This finding was corroborated by another study in which hip abductor strength was not correlated with peak hip adduction during running (Baggaley et al., 2015). Taken together, the results of the previous research, along with the current data, do not support the idea that isometric strength is the key determinant of frontal plane pelvic movements. Apart from the muscle strength factor, the activation pattern of the muscles surrounding the pelvis and thigh could play a role in controlling the frontal plane pelvic movement. These ideas are discussed in more detail below.

6.7.2 A framework to explain increased pelvic drop during running

To date, there are no complete datasets which have reported on the association between CPD measured during the stance phase of running and the EMG activity of lower limb muscles, particularly the adductor muscles. Therefore, this is the first study to report on this potential link. A window of muscle activity immediately preceding the point of peak pelvic drop was determined for each individual as muscle activation during this period will determine, to a large degree, the forces applied at the hip joint and therefore the resulting pelvic and hip motions during early stance. During running, an external flexion moment is developed at the hip joint during the first half of the stance phase. This flexion moment is counterbalanced by the action of the GMax, the hamstrings and AM. The data of the current study suggests that some individuals may utilise a different muscle synergy to generate this hip extension moment, by increasing the relative contribution of the AM and decreasing the relative contribution of the hamstrings and GMax muscles.

There were some side-to-side differences in the correlations observed between CPD and muscle activation (Figure 6- 4). However, importantly, for AM, GMax and MHam, the

results were consistent. Specifically, CPD showed a significant strong positive correlation with the activity of AM and a significant negative correlation present with GMax and MHam. Such correlations may indicate that lower activity in the hip extensor muscles (GMax and MHam) is substituted by the higher activity of AM in attempt to stabilise the pelvis and maintain the hip extension force needed during the stance phase. The AM muscle is a very important hip extensor and has the capacity to generate approximately 70% of the force produced by the GMax (Arnold et al., 2010). For example, at 90° hip flexion, the AM has a larger moment arm for hip extension compared the main hip extensor muscle (Nemeth & Ohlsen, 1985). However, importantly, as well as acting as a hip extensor, AM also acts as an adductor. Therefore, if the muscle synergy between the hip extensors changes so that a higher contribution is provided by AM, then this could act to destabilise the pelvis. This idea is supported by the relatively strong correlation between AM activity and CPD, which observed on both the left and the right sides.

An interesting finding was the strong positive correlation between CPD and the sagittal plane moment, on both sides. This relationship could have been the result of additional AM activity, which was associated with increased CPD. Although GMax and MHam displayed negative correlation with CPD, it is possible that the activity of these two muscles did not sufficiently decrease to reduce the sagittal plane moment at the point of peak CPD and this led to a strong association between CPD and increased sagittal plane hip moment.

Another interesting finding of the study was the opposite correlations between CPD and GMed activity between the different sides. On the right side, the activity of the GMed was negatively correlated with CPD and this suggest that there was no a corresponding increase in GMed to offset the increase in AM, which was associated with increased CPD. Conversely, on the left side, there was a positive correlation between CPD and GMed, which might

indicate that increased AM activity was counteracted by increased GMed activity in order to maintain pelvic stability. This idea could explain the positive correlation between the frontal plane (abduction) moment and CPD which was only observed on the left side. Taken together these different findings between the left and the right could indicate that GMed is not always able to counteract increased activity in the adductor muscles and this supports the idea that a broader understanding of hip muscle coordination and muscle synergy is required if pelvic drop is to be fully understood.

The data presented in this chapter support the idea that an over-activation of the adductor muscles (primarily the AM) could increase CPD. Furthermore, it would appear that increases in AM occur alongside decreases in GMax and MHam activity, suggesting that altered hip extensor muscle synergy could underlie CPD. Furthermore, the data also suggest that activity of GMed alone may not be adequate to counteract increased adductor activity and therefore control pelvic drop. These ideas indicate the need for a reconsideration of current clinical thinking in this area. Further research now is needed to explore synergetic control of the pelvic motion rather than the strength of the abductor muscles.

6.8 Limitations

One of the limitations of this study was the use of the hand-held dynamometry in measuring the strength of the hip abductors and adductors. A number of aspects to the protocol were optimised for the current study in order to maximise the validity of the strength measurements. This included using consistent testing positions and giving consistent directions during testing to limit the potential influence of verbal feedback on subject motivation. In addition, adequate recovery time between trials was given in order to avoid error due to participant exhaustion. This protocol matches that used in other previous research (Willy & Davis, 2011).

Another limitation of this study that an estimate of the EMD used had been derived from a different muscle and task and applied to the data in this thesis. This decision was made because no data were available on EMD from the adductor muscles and/or during running. Furthermore, visual inspection of the data showed that, after adjusting for EMD, the peak of GMed activity occurred at precisely the end of the EMG time window, see figure below (Figure 6- 6). This provides confidence that the decision to estimate the EMD from other muscles/other activities provides an appropriate estimate of the time delay for this data analysis.
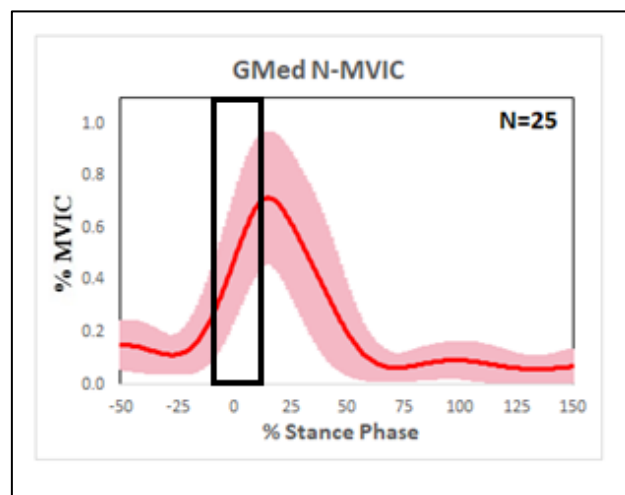
Figure 6- 6: Example of window shift for the glutes medius muscle when normalised to MVIC.

6.9 Conclusion

This study aimed to investigate the association between the frontal plane pelvic movement, muscle strength, hip moments and the EMG activity of the lower limb muscles during the early stance phase of running. The results of this study do not support the current thinking that isometric abductor strength will dictate frontal plane pelvic movements. Instead, the findings highlight the important functional role that the adductor muscles, particularly AM, play in determining frontal plane pelvic mechanics during running. Until now, little attention has been given to these muscles despite the fact that increased pelvic drop has been linked to

a range of different running-related injuries. The findings of the current study will motivate future research which should investigate whether rehabilitation programmes should incorporate muscle coordination retraining to restore the balanced activity of the muscles surrounding the pelvis and to understand whether this could lead to an improvement in frontal plane pelvic mechanics.

Chapter 7: Summary, Conclusions, Global Limitations and Recommendations for Future Work

The aim of the work presented in this thesis was to establish a valid and reliable method for measuring the EMG activity for the adductor muscles. The profile of activity for this group of muscles was then described during running. With this methodology and understanding of function, an investigation was carried out to understand the extent to which patterns of frontal plane motions in running, previously linked to pathology, could be linked to adductor function. In this chapter, a summary and key finding of the studies are presented. In addition, global limitations are discussed and a number of recommendations for the future work are suggested.

7.1 Summary

Following a systematic review of the literature surrounding the EMG for the adductor muscles during dynamic tasks, it was found that this muscle group should be considered as one of the most important muscle group of lower limb muscles. It represents approximately 22.5% of the total muscle mass of the lower extremity. In addition, the adductor group has the capacity to generate force both in frontal and sagittal planes. However, only a small number of studies have explored EMG activity for the adductor muscles during gait. The reason behind this paucity of research investigating the function of the adductors during ambulatory tasks could be due to the difficulties in measuring the EMG for this muscle group. Therefore, the first part of this thesis focused on developing a standard and reliable method for measuring the EMG activity for the adductor muscles. Following a description of the typical activation profiles of the lower limb muscles, their association to frontal plane pelvic movement in running was investigated. In total, the thesis comprises four studies:

Study 1: There were three separate but complementary aims to this study. In the first experiment, the relative movement of the adductor muscles under the skin was quantified at different hip joint angles. In the second experiment, the adductor muscles' positions were quantified during a ramped isometric contraction to identify the effect of muscle contraction on the relative position of the muscle. In the third experiment, an investigation was carried out into the relationship between adductor torque and the magnitude of the adductor EMG signals during ramped isometric contraction. Ten participants were recruited for this study.

Study 2: The first aim of this study was to investigate the degree of consistency between EMG measurements (from lower limb muscles) collected during overground walking and running on two different occasions. Whereas, the second complementary aim was to compare the different normalisation techniques of muscle activation during walking and running from the relevant methods available in the literature. The final aim was to assess the between-day reliability of measuring 3D biomechanical variables during walking and running activities. Ten participants were recruited for this study.

Study 3: This study aimed to describe the typical EMG profile for the adductor muscles during running as well as their inter-subject variability. In addition, the study aimed to describe the typical EMG profile for the major lower limb muscles collected during running as well as the associated inter-subject variability. Twenty-five participants were recruited for this study plus ten participants were added from the reliability study to describe five more muscles.

Study 4:

This study aimed to investigate the association between the frontal plane pelvic movement and the muscle strength, hip moments and the EMG activity of the lower limb muscles during the early stance phase of running.

In the first study, the results for the first aim displayed at least 5 mm from each side of the electrode borders for the three adductor muscles throughout the different hip extension/flexion movement. Such distances indicate the possible dimensions away from the neighbouring muscles. This confirms the location of the muscle within the electrode detection volume. As the surface electrodes were always placed centrally in a longitudinal orientation along the muscle fibres, there is minimal chance of picking up signals from the neighbouring muscles. Similar to the findings of the first aim, the ramped protocol of isometric muscle contraction examined in the second aim did not affect the location of the muscle under the electrode. The results showed at least 5 mm from each side of the electrode borders for the AL during the ramped protocol of isometric contraction. This distance was decreased to approximately 3 mm in AM and Gr muscles (observed only during the high percent MVIC). These data show that the electrode position relative to the underlying muscles did not change significantly during different levels of isometric contraction, demonstrating the minimal chance of cross talk from neighbouring muscles at the different levels of muscle contraction.

For the third aim of the first study, the EMG-torque relationship of the adductor muscles was investigated using surface EMG electrodes. In all tested hip adductor muscles, a statistically significant correlation was found between torque and the corresponding EMG activity. Therefore, the results for the third aim demonstrated a strong EMG-force relation.

In the second study, a between-days reliability study was conducted to investigate the degree of consistency between EMG measurements for lower limb muscles and the 3D kinematic variables during both walking and running tasks. For the first aim, the results of the tested muscles during both walking and running revealed good to excellent CMC values and low values of the SEM. For the second aim, the reliability analysis showed that the mean and peak of the dynamic trails as normalisation methods were the most reliable normalisation techniques across all tested muscles. These two normalisation methods reduced inter-individual variability of gait EMG signals compared to the MVIC method. However, the mean and peak normalisation methods decrease variability they also remove important information on the overall level of activation and therefore may hide actual differences between subjects. Thus, it was proposed the MVIC as the best method if an overall measure of the level of activation is required. For the third aim of the study, the sagittal and frontal kinematic variables showed excellent CMC values and low SEM values in both walking and running tasks.

In the third study, the three adductor muscles displayed a different pattern of activity during running. The AL peaks around toe-off, while Gr and the AM peaks around the mid-stance and the initial contact, respectively. However, the general pattern is present throughout the stance, there is a relatively high level of inter-subject variability for the three adductors. All other lower limb muscles peak around the initial contact apart from MGastro and LGastro that peak around the late stance. In general, the inter-subject variability of the adductors was higher than for other muscles. It was suggested that this is due to the synergistic nature of this muscle group, which can act in both the frontal and sagittal planes. Specifically, it is possible that different subjects used these muscles to varying extents to assist the muscles involved in generated sagittal plane moments.

In the fourth study, the most important findings of this study were that, on both sides, increased pelvic drop appeared to be strongly associated with increased AM activity and decreased GMax and hamstring activities. The findings suggest that some individuals may utilise a different muscle synergy to generate the hip extension moment, by increasing the relative contribution of the AM and decreasing the relative contribution of the MHam and GMax muscles. In summary, the findings of the current study show that increased pelvic drop is strongly associated with increased activity in the AM. This altered synergy will destabilise the pelvis and lead to altered frontal plane kinematics.

7.2 Conclusion

The work undertaken in this thesis has expanded the knowledge on the use of EMG for measuring the activity of the adductor muscles using surface electrodes. EMG data collected using the proposed protocol is unlikely to be affected by cross talk and that the amplitude of the signals is likely to reflect the level of force produced during dynamic movement. This thesis has also demonstrated that the proposed EMG protocol displayed good to excellent between-days repeatability. The thesis recommends the use of MVIC in normalising EMG amplitudes when exploring the variation between subjects, although the reliability was slightly lower than with other normalisation techniques, it was still good and the MVIC methods retain important information on the overall level of activation of muscles during running and walking. Finally, the thesis highlights the important role of the adductor group, especially, the AM in the frontal plane movement such as the pelvic drop. This finding motivates future work on muscle coordination retraining as a new clinical strategy for improving frontal plane mechanics during running.

7.3 Global limitations

The generalisability of the current project results is subjected to a number of limitations. Firstly, the cohort studied needs to be discussed. This cohort was limited to young, fit, lean male runners who were physically active between the ages of 18–40 and who ran at least 10-15 miles per week. This cohort will limit the generalisability of the findings and therefore, further research is required on female runners, elite runners with weekly higher mileages and also more sedentary cohorts. Moreover, participants, in the current project, wore standard training shoes on a standardised running surface. Therefore, it is not clear whether similar findings would be obtained when running on different surfaces with different footwear types, such as studded boots on grass or football trainers on AstroTurf. Future studies should be conducted while subjects wear their own training shoes and on a range of different running surfaces.

The second set of limitations relates to the study and measurement protocol. This thesis was novel as it used ultrasound to map out the position of a muscle group (the adductors), which until now had received minimal attention in the running literature. Nevertheless, it was possible to demonstrate good repeatability of EMG data collected from this muscle group, there was some day-to-day variability which might affect the results. Furthermore, it is not clear to what extent the findings can be generalised from a laboratory setting to real-world environments. For example, the current study was limited to one speed as exploring the effect of running at different speeds on the muscle activation pattern was beyond the scope of this thesis. Therefore, further studies are needed to understand whether the link between adductors function and CPD is consistent across different speeds. With ongoing technological evolutions, such as ambulatory EMG systems, it would be possible to extend the current

investigation into more ecologically valid evaluations of loading and injury risk in actual sports environments and training sessions.

7.4 Recommendations for future work

The findings of this thesis and the subsequent discussion raise several questions, which could be investigated in future research. First, following the results of the study presented in chapter 3, it is recommended to use the same protocol in locating and measuring the activity of the adductor muscles. Further research is required to monitor the oscillatory movement of the thigh during running. In addition, the EMG-force relation needs to be explored during dynamic contractions such as concentric and eccentric contractions. Following the results of the reliability study described in chapter 4, it is recommended to use the MVIC normalisation method, as it provides useful information about the actual activity of the muscle in relation to its maximum capacity. In addition, acceleration during a dynamic task plays a very important role in obtaining a good to excellent reliable EMG measures across testing sessions. Further research into different populations, including a variable level of runners using different speed, would be useful to ascertain the results of this study. The finding of the study presented in chapter 6 motivates future clinical trials which could focus on both muscle coordination retraining aiming at decreasing the incidence of running-related injuries. Using biofeedback could be a good example for gait retraining which optimises the coordination patterns during running; thus, improving the frontal plane pelvic movements.

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Appendices

Appendix I

Search strategy for the adductor muscles

Cochrane Pubmed(medline) (from, Academic search premier (ENSCO): CINAHL, Medline, Sportdiscus, and Library information science and technology abstract, Web of knowledge.

1. electromyogra* or EMG*or * EMG

2. hip adduct*

3. Limiter: human and English

Identified studies from search strategy:

Medline = 155

Sportdiscus 103

Academic search premier =77

CINAHL=63

Web of knowledge 200

Total n= 598.

Papers excluded: duplications n= 291.

Papers retrieved for examination: n= 307.

Papers excluded: Irrelevance n= 188.

Neurological conditions n=49

Papers retrieved for examination: n= 70

Participant's condition

- **Musculoskeletal condition n.= 8.**

  Acetabular labral tears (2), groin strain (1), hip pain (1), knee OA (1), PFPS (3).

- **Healthy n.= 62.** (some articles measured the EMG activity during more than one task).

  Cutting (2), cycling (5), football tasks (4), golf (1), hip adduction ex's (3), horseback riding (1), isokinetic testing (2), landing (2), leg press (1), rapid leg flexion (2), running (8), skating (1), SLR (3), squatting (11), stair ambulation (1), static tests (7), taekwondo (1), jump (1), walking (7), weight bearing (1), weight bearing (1), and weight lifting (2)

Type of electrode used during the measurement of healthy subjects:

- **Fine wire n.=7.**

  Gracilis (1), pectineus (1), adductor brevis (1), adductor longus (3), and adductor magnus (5).

- **Surface electrodes n.=65.** (2 articles use both types in their studies).

  Not mentioned (17), gracilis (6), pectineus (0), adductor brevis (0), adductor longus (34), and adductor magnus (17).

  "Note: the number of detailed descriptions for both types are not matched with the total number of each type. This is because some articles measured the EMG activity of more than one muscle".

Appendix II

The approval letter of the study

University of **Salford**
MANCHESTER

Research, Innovation and Academic
Engagement Ethical Approval Panel

Research Centres Support Team
G0.3 Joule House
University of Salford
M5 4WT

T +44(0)161 295 2280

www.salford.ac.uk/

21 March 2017

Dear Walaa,

RE: ETHICS APPLICATION–HSR1617-82–'Using ultrasound to monitor relative position of the adductor muscles in order to inform EMG electrode placement.'

Based on the information you provided I am pleased to inform you that application HSR1617-82 has been approved.

If there are any changes to the project and/or its methodology, then please inform the Panel as soon as possible by contacting Health-ResearchEthics@salford.ac.uk

Yours sincerely,

Sue McAndrew
Chair of the Research Ethics Panel

Appendix III

A Randomization Plan
from
http://www.randomization.com

1. _____
   - MVIC extension
   - Left
   - Right
   - MVIC adduction

2. _____
   - Left
   - MVIC extension
   - Right
   - MVIC adduction

3. _____
   - MVIC adduction
   - MVIC extension
   - Left
   - Right

4. _____
   - Right
   - MVIC extension
   - Left
   - MVIC adduction

5. _____
   - MVIC adduction
   - Left
   - Right
   - MVIC extension

6. _____
   - Right
   - MVIC adduction
   - MVIC extension
   - Left

7. _____
   - Right
   - Left
   - MVIC extension
   - MVIC adduction

8. _____
   - MVIC extension
   - MVIC adduction
   - Right
   - Left

9. _____
   - MVIC extension
   - Right
   - MVIC adduction
   - Left

10. _____
   - Left
   - MVIC adduction
   - Right
   - MVIC extension

Appendix IV

The approval letter of the study

University of
Salford
MANCHESTER

Research, Innovation and Academic
Engagement Ethical Approval Panel

Research Centres Support Team
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www.salford.ac.uk/

24 June 2016

Dear Walaa,

RE: ETHICS APPLICATION HSCR 16-44 –Repeatability of EMG during running

Based on the information you provided, I am pleased to inform you that application HSCR16-44 has been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible by contacting Health-ResearchEthics@salford.ac.uk

Yours sincerely,

Sue McAndrew
Chair of the Research Ethics Panel

Appendix V

A Randomization Plan

from

http://www.randomization.com

1. _____
   - ○ R
   - ○ L

2. _____
   - ○ L
   - ○ R

3. _____
   - ○ L
   - ○ R

4. _____
   - ○ L
   - ○ R

5. _____
   - ○ R
   - ○ L

6. _____
   - ○ L
   - ○ R

7. _____
   - ○ R
   - ○ L

8. _____
   - ○ R
   - ○ L

9. _____
   - ○ L
   - ○ R

10. _____
   - ○ R
   - ○ L

10 subjects randomized into 1 block
To reproduce this plan, use the seed 27131
Randomization plan created on 6/8/2016, 6:11:57 PM

1/1

Appendix VI

<div align="center">

Pilot study

</div>

Aim:

This pilot aimed at determining the maximum activation levels of the hip adductor muscles in two different clinical tests in order to be used in normalisations of EMG amplitudes obtained from dynamic tasks.

Data processing:

The data were exported as a C3D to Matlab for processing with custom written software in Matlab. There are three steps to process the raw EMG data: first step is used a high pass filter for walking (20Hz) and running (30Hz) to remove movement artefacts and noise. Second step is rectification and envelope detection which makes all signals positives. Final step is a low filter (6Hz) which is used by (Hubley-Kozey et al., 2006; Hubley-Kozey et al., 2010) to create a linear envelope. Following EMG processing, the data were exported to Microsoft Excel 2016 sheet to obtain final result.

Statistical analysis:

The within subject paired t-tests were used to determine the adductor muscle's maximum voluntary isometric contraction (MVIC) at both hip angles (0° and 45°).

Recap of results:

For high bass filter of 20Hz, the filter used for walking dataset, there was a significant decrease in MVIC activity for AL muscle at 45° hip flexion angle ($p = 0.036$). Additionally, the AM demonstrated significant greatest activation at 45° hip flexion angle ($p = 0.000$).

Similarly, the Gr. demonstrated greatest (non-significant) activation at 45º hip flexion angle (table 1).

Table 1: Summary of the pilot study results for the MVIC amplitudes using high pass filter of 20Hz (for walking activity). SD: Standard Deviation; AL: Adductor longus; AM: Adductor magnus; Gr: Gracilis.

*Significant < 0.05

Adductor muscles	Mean	SD	P value	Cohen's d
AL 0º	365.62	119.44	0.028*	0.9
AL 45º	267.79	102.69		
Gr 0º	177.30	94.95	0.239	0.6
Gr45º	218.13	53.39		
AM 0º	72.20	26.86	0.005*	1.4
AM 45º	127.44	53.07		

Similarly, for high bass filter of 30Hz, the filter used for running dataset, the MVC activity of AL muscle at 45º hip flexion angle of was significantly lower than that at 0º hip flexion angle. In addition, AM demonstrated significant greatest activation in hip 45 ($p = 0.000$), as well as the GR demonstrated greatest (non-significant) activation in hip 45 (table 2).

Table 2: Summary of the pilot study results for the MVIC amplitudes using high pass filter of 30Hz (for running activity). SD: Standard Deviation; AL: Adductor longus; AM: Adductor magnus; Gr: Gracilis.

Adductor muscles	Mean	SD	P value	Cohen's d
AL 0º	359.21	94.44	0.044*	1.3
AL 45º	236.14	93.05		
Gr 0º	161.79	52.56	0.614	0.2
Gr45º	171.58	46.15		
AM 0º	85.59	15.50	0.000*	2.8
AM 45º	145.06	26.96		

*Significant < 0.05