Assessment of Plantarflexor Function during a Stretch-Shortening Cycle Task

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A thesis submitted to the University of Limerick in fulfilment of the requirements of the degree of Doctor of Philosophy

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Dedicated to a four-legged friend
Abstract

Title: Assessment of Plantarflexor Function during a Stretch-Shortening Cycle Task

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The plantarflexors are important due to their role in locomotion and stiffness control, high prevalence of injury, link to knee stability and anterior cruciate ligament injury. Current methods of assessing the plantarflexors are limited primarily by issues of validity and reliability. The aims of this thesis were to develop a method of measuring plantarflexor function in a dynamic yet controlled stretch-shortening cycle (SSC) task and to assess plantarflexor muscle-tendon unit (MTU) function in healthy young adults using this method.

Use of an adapted force sledge was found to be a highly reliable method to determine plantarflexor force, force development and reactive strength capabilities during a fast SSC task (single ICC >0.85, average ICC >0.95). Using this method, moderate between-limb differences in measures of force, force development and reactive strength index of up to 44.6% were observed in 21 young healthy adults. These differences are much larger than the 15% upper limit previously suggested for healthy limbs. Differences in plantarflexor force, force development and mechanical behaviour of the plantarflexors were observed due to gender in a group of 34 age and training-matched individuals. While some differences were removed with normalisation to muscle volume, differences in peak force and power remained, which suggests underlying differences in the MTU which may affect injury risk or training response.

To provide further insight into underlying muscle-tendon behaviour, a motion analysis-based method of determining centre of pressure position was developed. This was shown to be reasonably accurate in location of the centre of pressure during two-legged hopping at frequencies greater than 1.5 Hz. Incorporating this method with the adapted force sledge and ultrasound, the main determining factors of stiffness during this task were found to be the rate of peak force development, rate of force development in the first 30 ms and contact time. Measures previously suggested to be associated with tendon stiffness, namely force development time and elastic charge time, were found to have no relationship with Achilles tendon stiffness during the stretch-shortening cycle task. Similarly, no relationship was shown between Achilles tendon stiffness and reactive strength index. Instead, reactive strength index was determined by normalised peak eccentric power, concentric work and peak concentric ankle joint power during the impact.

In summary, an adapted force sledge can be used to reliably assess plantarflexor MTU function during a SSC task and investigate several research questions related to the function of this particular muscle group. It can also be combined with motion analysis to determine joint kinetics. The results of this research are important for both practitioners and researchers for understanding healthy plantarflexor function during a SSC task. This work has implications for the design and selection of appropriate test protocols and understanding what is being measured, both for establishing normative data in a valid setting and for understanding the range of measures that may be observed in healthy young active adults.
Declaration

I hereby declare that the work contained in this thesis is my own, and was completed with counsel of my supervisor, Dr. Andrew J Harrison of the Department of Physical Education and Sport Sciences, University of Limerick. The work has not been submitted to any other university or higher education institution, or for any other academic award with this university.

__________________________       ________________________
Laura-Anne M Furlong               Dr. Andrew J Harrison
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<th>Description</th>
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<tbody>
<tr>
<td>%CT</td>
<td>Percentage of contact time</td>
</tr>
<tr>
<td>11RM</td>
<td>11 repetition maximum</td>
</tr>
<tr>
<td>1MTP</td>
<td>1\textsuperscript{st} metatarsophalangeal joint</td>
</tr>
<tr>
<td>3D</td>
<td>Three-dimensional</td>
</tr>
<tr>
<td>3MTP</td>
<td>3\textsuperscript{rd} metatarsophalangeal joint</td>
</tr>
<tr>
<td>5MTP</td>
<td>5\textsuperscript{th} metatarsophalangeal joint</td>
</tr>
<tr>
<td>95% CI</td>
<td>95% confidence interval</td>
</tr>
<tr>
<td>ACL</td>
<td>Anterior cruciate ligament</td>
</tr>
<tr>
<td>AP</td>
<td>Anteroposterior</td>
</tr>
<tr>
<td>AT</td>
<td>Achilles tendon</td>
</tr>
<tr>
<td>CE</td>
<td>Contractile element</td>
</tr>
<tr>
<td>CMJ</td>
<td>Countermovement jump</td>
</tr>
<tr>
<td>CoP</td>
<td>Centre of pressure</td>
</tr>
<tr>
<td>CoV</td>
<td>Coefficient of variation</td>
</tr>
<tr>
<td>CT</td>
<td>Contact time</td>
</tr>
<tr>
<td>ECT</td>
<td>Elastic charge time</td>
</tr>
<tr>
<td>EMD</td>
<td>Electromechanical delay</td>
</tr>
<tr>
<td>EMG</td>
<td>Electromyography</td>
</tr>
<tr>
<td>ES</td>
<td>Effect size</td>
</tr>
<tr>
<td>FDT</td>
<td>Force development time</td>
</tr>
<tr>
<td>FP</td>
<td>Force plate</td>
</tr>
<tr>
<td>F_p</td>
<td>Peak force</td>
</tr>
<tr>
<td>F_{P,NORM}</td>
<td>Peak force (normalised to muscle volume)</td>
</tr>
<tr>
<td>fSSC</td>
<td>Fast stretch-shortening cycle</td>
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FT  Flight time
GR  Groove
ICC  Intraclass correlation coefficient
ITB  Iliotibial band
K\textsubscript{ANK}  Ankle stiffness
K\textsubscript{ANK(NORM)}  Ankle stiffness (normalised to muscle volume)
K\textsubscript{AT}  Achilles tendon stiffness
KJC  Knee joint centre
K\textsubscript{PF}  Plantarflexor stiffness
K\textsubscript{PF(NORM)}  Plantarflexor stiffness (normalised to muscle volume)
MALL  Malleolus
MF  Mid-foot
ML  Mediolateral
MTJ  Muscle-tendon junction
MTP  Metatarsophalangeal
MTU  Muscle-tendon unit
MVC  Maximal voluntary contraction
P\textsubscript{A\textsubscript{CONC}}  Peak ankle joint power (concentric phase)
P\textsubscript{A\textsubscript{CONC(NORM)}}  Peak ankle joint power (concentric phase, normalised to muscle volume)
P\textsubscript{AECC}  Peak ankle joint power (eccentric phase)
P\textsubscript{AECC(NORM)}  Peak ankle joint power (eccentric phase, normalised to muscle volume)
P\textsubscript{CONC}  Peak power (concentric phase)
PCSA  Physiological cross-sectional area
PE  Parallel elastic element
P\textsubscript{ECC}  Peak power (eccentric phase)
PS\textsubscript{CONC}  Peak plantarflexor power (concentric phase)
PS<sub>CONC(NORM)</sub>  Peak plantarflexor power (concentric phase, normalised to muscle volume)

PS<sub>ECC</sub>  Peak plantarflexor power (eccentric phase)

PS<sub>ECC(NORM)</sub>  Peak plantarflexor power (eccentric phase, normalised to muscle volume)

RFD<sub>0-30</sub>  Rate of force development in first 30 ms

RFD<sub>0-30(NORM)</sub>  Rate of force development in first 30 ms (normalised to muscle volume)

RFD<sub>0-50</sub>  Rate of force development in first 50 ms

RFD<sub>0-50(NORM)</sub>  Rate of force development in first 50 ms (normalised to muscle volume)

RMSD  Root mean square difference

RPFD  Rate of peak force development

RPFD<sub>NORM</sub>  Rate of peak force development (normalised to muscle volume)

RSI  Reactive strength index

SE  Series elastic element

SSC  Stretch-shortening cycle

WA<sub>CONC</sub>  Ankle work done (concentric phase)

WA<sub>CONC(NORM)</sub>  Ankle work done (concentric phase, normalised to muscle volume)

WA<sub>ECC</sub>  Ankle work done (eccentric phase)

WA<sub>ECC(NORM)</sub>  Ankle work done (eccentric phase, normalised to muscle volume)

W<sub>CONC</sub>  Work done (concentric phase)

W<sub>ECC</sub>  Work done (eccentric phase)

WS<sub>CONC</sub>  Plantarflexor work done (concentric phase)

WS<sub>CONC(NORM)</sub>  Plantarflexor work done (concentric phase, normalised to muscle volume)

WS<sub>ECC</sub>  Plantarflexor work done (eccentric phase)

WS<sub>ECC(NORM)</sub>  Plantarflexor work done (eccentric phase, normalised to muscle volume)
Submissions and publications arising from and related to the thesis

Peer-reviewed journal articles


Journal articles in review


Peer-reviewed international conference proceedings (4 page paper)


Furlong, L.A.M. and Harrison, A.J. (2013) 'Leg dominance and plantarflexor stretch-
shortening cycle function', in Shiang, T.-Y., Ho, W.-H. and Chenfu, P. (eds.) Scientific
Proceedings of the 31st International Conference on Biomechanics in Sports, Taipei, Taiwan, July 7-11, National University of Taiwan: International Society of Biomechanics in Sports.

Proceedings of the 31st International Conference on Biomechanics in Sports, Taipei, Taiwan, July 7-11, National University of Taiwan: International Society of Biomechanics in Sports.

of motion during dynamic tasks?’, in Shiang, T.-Y., Ho, W.-H. and Chenfu, P. (eds.) Scientific
Proceedings of the 31st International Conference on Biomechanics in Sports, Taipei, Taiwan, July 7-11, National University of Taiwan: International Society of Biomechanics in Sports.

during cyclical loading of the triceps surae’, Portuguese Journal of Sport Sciences, 11 (S2), 887-
890.

Peer-reviewed national conference proceedings (300 word abstract)

Chapter 1: Introduction
1.1 Background

The primary ankle plantarflexor muscles are gastrocnemius, soleus and plantaris which are located in the posterior compartment of the leg (O’Brien 2005). The muscle group is an important force producer in running, particularly in the acceleration and maximal velocity phases of the sprint (Johnson and Buckley 2001; Bezodis et al. 2008). The bipennate, biarticular gastrocnemius is the largest and most superficial of the three muscles and acts as both ankle plantarflexor and knee flexor. Its primary role is that of a rapid force producer (Edgerton et al. 1975; Harper and Trappe 2008) but it may also act to reduce risk of ankle eversion injury (Antonios and Adds 2008). Soleus is primarily composed of slow-twitch fibres and has an important role to play in postural control and stability (Kuitunen et al. 2002b; O’Brien 2005).

As the tendon of gastrocnemius and soleus, the Achilles tendon is the largest and strongest tendon in the human body (O’Brien 2005). The mechanical characteristics of this tendon have been shown to be related to injury (Mahieu et al. 2006; Lichtwark et al. 2009), affect movement efficiency and economy (Lichtwark and Wilson 2005a; 2007; Stafilidis and Arampatzis 2007; Lichtwark and Wilson 2008; Scholz et al. 2008) and affect performance of dynamic tasks (Lichtwark and Wilson 2005a; Albracht and Arampatzis 2006; Karamanidis and Arampatzis 2006; Lichtwark and Wilson 2006; Barrett and Lichtwark 2008; Lichtwark and Barclay 2010). In addition to this, tendon has also been shown to be advantageous in development of bipedal gait (Bramble and Lieberman 2004), decreasing risk of soft tissue damage (McHugh et al. 1999) and providing a store of elastic energy (Biewener and Baudinette 1995).

The plantarflexors and Achilles tendon are two of the most common sites of overuse injury, the reasons for which are still unclear (Magnusson et al. 2010). Approximately 50% of runners will experience injury each year, with 25% injured at any given time (Fields et al. 2010). The percentage of running injuries occurring in the calf and ankle area ranges from 23.3% (Taunton et al. 2002) to 57.7% (Maughan and Miller 1983; van Middelkoop et al. 2008; Buist et al. 2009; Chang et al. 2013), dependent on injury definition. Despite the importance of this muscle group and its high rate of injury, there are a number of issues associated with the assessment of its function.

Current methods of assessing the plantarflexors are limited primarily by issues of validity and reliability. Studies of in vivo plantarflexor function in humans involve dorsiflexion-
plantarflexion movements on a force plate (Kubo et al. 2000c) or dynamometer (Svantesson et al. 1991; Kubo et al. 2005a). The dynamometer improves control but lacks ecological validity for real-life activities such as hopping and running due to kinematic differences, lack of impact events and altered loading and rate of force development. Reliability of ankle stiffness during a force-plate based hopping protocol has been shown to be poor (Joseph et al. 2013), but measures of lower limb function using force sledges have been shown to be highly reliable (Flanagan and Harrison 2007). This method has previously been used to study the entire lower limb, but has not been used previously to investigate isolated muscle group function. Oscillation and quick-release techniques are often used to study the mechanical properties of the MTU. The oscillation technique involves modelling the limb as a linear damped-spring system. Oscillations in the force data are measured and input into formulae to determine musculoarticular stiffness. Low levels of perturbation suggest high levels of stiffness, and vice versa (Ditroilo et al. 2011). The quick-release technique measures the stiffness of the series elastic element of the MTU (Goubel and Pertuzon 1973). This method requires participants to perform an isometric plantarflexion contraction at a certain percentage of maximal voluntary contraction against a plate which is then suddenly released. Stiffness is calculated as change in joint moment divided by angular displacement (similar to joint stiffness) and related to the isometric torque of the preceding 20 ms with the stiffness-torque relationship considered an index of MTU stiffness. Reliability is established for both methods (Murphy 2003; McLachlan et al. 2006b; Lambertz et al. 2008) with construct validity established for the oscillation technique (Walshe et al. 1996). Neither method provides a direct measure of the stiffness of the tendon itself as this requires measurement of both force and tendon elongation.

The mechanical properties of the lower limb have been shown to vary with task demands (Ferris and Farley 1997; Farley and Morgenroth 1999; Arampatzis et al. 2001; Günther and Blickhan 2002; Kuitunen et al. 2002b; Karamanidis et al. 2006), age (Lambertz et al. 2003), training status (Harrison et al. 2004; Hobara et al. 2008), gender (Eiling et al. 2007) and level of fatigue (Comyns et al. 2011). Measures at limb level are composite measures, a quasi-measure of the behaviour of the joints, ligaments, tendons and connective tissues of the entire lower limb (Latash and Zatsiorsky 1993). Understanding what happens at joint and tissue level may greatly enhance our understanding of complex interactions during locomotion, but to date most work at this level has used dynamometry. There is a need to improve existing methods of measuring plantarflexor function, as they currently either
assess the muscle-tendon unit in a mode of contraction that is not valid for dynamic everyday activities, or lack measurement control. Development of a method that could assess this muscle group in a dynamic yet controlled stretch-shortening cycle test environment would allow us to obtain normative data for plantarflexor function in a variety of populations, may help to explain observed performance differences as well as help us understand the determining factors of the mechanical behaviour of this particular muscle group.

1.2 Thesis aims

1. To develop a method of measuring plantarflexor muscle-tendon function in a dynamic yet controlled stretch-shortening cycle test environment
2. To assess plantarflexor muscle-tendon function in healthy young adults using this method

1.3 Thesis objectives

1. To quantify the reliability and consistency of a force-sledge based method of measuring plantarflexor muscle-tendon function
2. To investigate differences in plantarflexor muscle-tendon function in healthy young adults due to limb preference
3. To investigate differences in plantarflexor muscle-tendon function in healthy young adults due to gender
4. To investigate the potential use of a motion-analysis based method of quantifying the centre of pressure during hopping
5. To use this method to determine the main factors influencing plantarflexor stiffness
6. To investigate if electromechanical delay (force development time and elastic charge time) or reactive strength index are related to the mechanical properties and loading of the plantarflexors

1.4 Thesis structure

This thesis contains a series of progressively linked studies describing the development and application of a force-sledge based method of assessing plantarflexor muscle-tendon function.
Chapter 1 provides a brief introduction to the research topic and outlines the primary aims of the thesis.

Chapter 2 presents a comprehensive review of literature related to the thesis, focusing on the importance of the plantarflexors in dynamic stretch-shortening cycle tasks such as hopping and running and existing methods of measuring muscle-tendon function and mechanics.

Chapter 3 describes an alternative method of quantifying plantarflexor function in a dynamic, controlled environment. This chapter reports the reliability and consistency of measures of plantarflexor force, force development and stretch-shortening cycle function using an adapted force sledge apparatus.

Chapter 4 examines differences in plantarflexor function in young healthy adults due to limb preference using the adapted force sledge reported in chapter 3. Measures of force, force development and stretch-shortening cycle function for preferred and non-preferred hopping limbs were compared. Results enhance our understanding of typical levels of asymmetry observed in healthy adults and may explain why differences in ankle kinematics and kinetics are observed in sprinting.

Chapter 5 examines differences in plantarflexor function in young healthy adults due to gender. Measures of force, force development, stretch-shortening cycle function and mechanical properties for a sample of age and training-matched males and females were compared. This chapter may help to explain, in part, performance differences observed in males and females and guide practitioners to develop practices to optimise plyometric performance and minimise injury risk.

Chapter 6 describes a motion analysis marker-based method of calculating centre of pressure as the adapted force sledge does not allow for use of a force plate during the task. Researchers have attempted to use alternate methods of calculating centre of pressure during dynamic tasks with varying success, mainly due to errors in spatial and temporal synchronisation. This study is the first to investigate the feasibility of calculating centre of pressure data using a motion analysis system, and may be of benefit to clinicians and researchers who wish to increase the mobility of acquiring centre of pressure data.

Chapter 7 incorporates the methods developed in previous chapters to investigate the factors that influence plantarflexor stiffness during a fast stretch-shortening cycle task. The
chapter combines the force sledge method of assessing force and force development properties of the plantarflexors and the marker-based method from chapter 6 with ultrasound to quantify ankle and tendon mechanics during the task. This chapter also investigates the use of force development time, elastic charge time and reactive strength index as surrogate measures of muscle-tendon unit mechanics.

Chapter 8 summarises the key findings of the thesis and their practical implications. The limitations of the thesis and directions for future study are also identified.
Chapter 2: Literature review
2.1 The muscle-tendon unit

The primary role of medial and lateral gastrocnemius, soleus, peroneus longus and brevis, tibialis posterior, flexor hallucis longus and flexor digitorum longus is to plantarflex (extend) the ankle joint (Handsfield et al. 2014). The main muscles discussed in this literature review will be the gastrocnemii and soleus as they are the largest, most dominant plantarflexors and also the most commonly reported muscles in the literature.

2.1.1 Muscle and tendon

The ground-breaking research into skeletal muscle function of Archibald Vivian Hill and Otto Fritz Meyerhof in the early 20th century resulted in their success in winning the Nobel Prize in Physiology or Medicine in 1922. Since then, numerous researchers have devoted hours to the study of this complex arrangement of numerous sarcomeres under voluntary control which both exerts force and produce mechanical work (Zatsiorsky and Prilutsky 2012). Muscles’ capabilities and behaviour in in vivo conditions underpins how and why we move the way we do (Albracht and Arampatzis 2006; Earp et al. 2010; Kuitunen et al. 2011; Chumanov et al. 2012), why we choose a particular speed of movement (Neptune and Sasaki 2005; Bartlett and Kram 2008; Farris and Sawicki 2012b), how and why we fatigue (Peltonen et al. 2010; Comyns et al. 2011) and why we get injured (Fleming et al. 2001; Brockett et al. 2004; Adachi et al. 2008; O’Sullivan et al. 2008). Its ability to repair dictates how long we are out of competition or if we can return to competition at all, while muscle composition can help explain why some make fast sprinters and others run marathons (Harrison et al. 2004; MacArthur and North 2004). Recent technological advances now allow us to image muscle and tendon behaviour in vivo during walking and running tasks (Lichtwark and Wilson 2005b; Lichtwark and Wilson 2006; Farris and Sawicki 2012b; Farris et al. 2012; Cronin and Lichtwark 2013), which provides even greater insight into how muscle functions.

To say the only purpose of tendon is to join muscle to bone is akin to saying Michelangelo was just a painter: while both statements are true, they are not anyway close to describing their subject’s importance (Paxton and Baar 2007). In humans, tendon decreases the risk of soft tissue damage (McHugh et al. 1999), enhances muscle efficiency (Biewener 1997), provides a store of elastic energy (Biewener and Baudinette 1995) and its mechanical
properties can differentiate between power and endurance athletes (Kubo et al. 2000a; 2000b).

Tendon is composed of crimped collagen (mainly type I) fibres aligned in one direction and a viscoelastic matrix of proteoglycans, glycoproteins and glycosaminoglycans (Kafka et al. 1995). The structural unit of collagen is tropocollagen, with five tropocollagens forming fibrils of 20-150 nm diameter. Achilles tendon fibrils range from 30-130 nm in diameter, with most between 50 and 90 nm. Several parallel fibrils form fascicles which are the smallest collagenous structure that can be mechanically tested. Fascicles are surrounded by endotenon, epitenon and paratenon with fluid found between the top two layers to reduce friction (O'Brien 1997). The stress-strain curve of non-mineralised tendon is typically S-shaped and can be divided into three regions: the toe region, linear elastic region and the yield and failure region (Figure 1). The regions appear due to the straightening of the crimp in the collagen fibres, stretching of the tendon collagen fibres and the irreversible destruction of the crystalline pattern within the fibres beyond the yield point. The appearance of mineralisation of the tendon appears to minimise the appearance of the toe region with tendon stress-strain curves with high levels of mineralisation looking like those of bone (Landis et al. 1995).

![Stress-strain curve for non-mineralised tendon showing the three main regions](Korhonen and Saarakkala 2011)

*Figure 1. Stress-strain curve for non-mineralised tendon showing the three main regions (Korhonen and Saarakkala 2011)*
As the area between the muscle and tendon, the myotendinous junction is the area of
greatest stress during transmission of force (Charvet et al. 2012). It is where elongation
occurs and is considered the growth plate of tendon due to the presence of cells that can
elongate and deposit rapidly. Golgi tendon organs and nerve receptors are also located here
to prevent over-stretching (O'Brien 1997).

2.1.2 The Hill muscle model and its relevance to study of the MTU

The Hill muscle model (Figure 2) proposes the MTU consists of a contractile element
(CE), a series elastic element (SE) and a parallel elastic element (PE). Each part of the
model refers to a specific muscle behaviour rather than anatomical structure, despite
certain structures acting more like one component than another. The interaction between
the three components helps to explain observed performance differences in a variety of
tasks.

\[
\text{Figure 2. The Hill muscle model. CE = contractile element, SE = series elastic element,}
\]
\[
\text{PE = parallel elastic element (Wikipedia 2014)}
\]

The CE of the Hill model refers to the element that converts nervous signals into force,
the magnitude of which is dependent on four key relationships: stimulation-activation,
force-activation, force-length and force-velocity. Muscle behaves as a CE in the MTU.

1. The stimulation-activation relationship refers to the relationship that exists between
the amount of calcium that floods into the transverse tubules immediately prior to
cross-bridge coupling (i.e. input into the process) and the response of the actin-myosin
complex to the influx, both of which are difficult to precisely quantify. Activation refers to the state where force can be produced (Caldwell 2004).

2. The force-activation relationship is a conceptual, linear relationship. This relationship suggests that when muscle is activated by 10%, 10% of potential force is produced, when activated by 20%, 20% of potential force is produced etc. Previous work has suggested this relationship is related to the stiffness of the muscle (Ford et al. 1981). The validity of this relationship is questionable due to the difficulty in establishing 100% of force capacity and previous work showing EMG during running to be much greater than that observed during a 100% ‘maximal voluntary contraction’ (Kyröläinen et al. 2005)

3. The force-velocity relationship is well established. It is expressed mathematically as

\[(P + a) = P_o (v + b)\]

where \(P\) and \(v\) represent CE force and velocity at a given instant in time, \(P_o\) = the force level the CE would attain at that instant if it were isometric and \(a\) and \(b\) are the muscle dynamic constants which Hill used to describe the constants of energy liberation (Hill 1938). This equation is only valid for isometric or concentric contraction velocities, with modifications required for eccentric contractions (Caldwell 2004). Inter- and intra-species constants differences exist, with different muscle types demonstrating different hyperbolas and intercepts on the force and velocity axes.

![Figure 3. Force-velocity relationship for skeletal muscle (Kreighbaum and Barthels 1996)](image-url)
4. The fourth relationship of this model is the muscle force-length relationship (Ramsey and Street 1940). This relationship is proposed to be a concave-up, single hyperbola, with peak isometric force proposed to occur at mid-muscle length which is also known as rest length. This relationship is shown graphically in Figure 4.

![Figure 4. Force-length relationship for skeletal muscle, showing ascending limb (a-b), plateau region (c-d) and descending limb (d-e) with corresponding filament overlap, adapted from Zatsiorsky and Prilutsky (2012)](image)

The SE refers to elastic structures that are in series with the force-generating machinery of muscle. The SE does not have force-producing capabilities of its own but rather is dependent on the CE to alter its elastic properties. Within the MTU, a certain amount of elasticity exists not only in the CE but also in the tendon, aponeurosis and other connective tissues present. These structures typically are in series with the CE so any force generated in the CE is also transmitted across these structures (Roy et al. 2003). It is possible to measure a number of mechanical properties of the SE, such as stiffness, strain and Young’s modulus.

Even when no force is produced by the CE, there is still resistance from the MTU to stretch which is not from the CE or SE; this resistance comes from the PE which is
typically associated with the fascia that surrounds and divides the plantarflexors. Similar to the SE, PE stiffness is normally non-linear. The force-tension relationship of the passive PE is different to that of the CE with low forces associated with low tension which then increases upwards in a curvilinear fashion. This is in contrast to the hyperbolic curve of the force-tension relationship of the CE (Caldwell 2004).

Understanding the characteristics of the three components of the MTU helps us understand the control of movement, as trade-offs occur between each component during different movements and different contraction types. For example, there appears to be a trade-off between SE stiffness and CE shortening velocity to maintain the MTU in a condition which optimises force production and minimises energetic costs. It was previously thought that at low forces the SE was very compliant and to maintain isometric contraction the CE had to contract at very high shortening velocities. The opposite has been shown in one-legged hopping where high SE stiffness and low CE shortening velocity was observed, which subsequently reduces the energetic cost of movement (Lichtwark and Wilson 2005b).

2.2 The plantarflexors

2.2.1 Anatomy

The bipennate, biarticular gastrocnemius is the most superficial plantarflexor muscle and acts as both ankle plantarflexor and knee flexor. It is composed of between 50 and 57% fast twitch fibres (Gollnick et al. 1973; Edgerton et al. 1975). Both the medial and lateral heads originate at the popliteal ligament and knee joint capsule with strong flat tendon attaching them to the femoral condyles. This tendon extends for a short distance on the posterior superficial surface of the muscle as aponeurosis. The two heads also have secondary individual origins. The longer and larger medial head originates at the popliteal surface of the femur behind the medial supracondylar line and the adductor tubercle, above the medial femoral condyle. The lateral head runs from the posterior part of the lateral surface of the lateral femoral condyle, above and behind the lateral epicondyle and from a portion of the linea aspera above the condyle. These two bellies merge and insert into dense aponeurosis anteriorly, blending into soleus to form the Achilles tendon (AT). The point where this occurs varies from person to person, and the tendon subsequently inserts
onto a smooth transverse area on the middle third of the posterior surface of calcaneus (Drake et al. 2005; O’Brien 2005).

Muscle volume refers to the cubic area of the muscle of interest, usually measured in cm$^3$. Physiological cross-sectional area (PCSA) refers to the muscle volume divided by fibre length, accounting for muscle density and pennation angle (Narici et al. 1996) and is proportional to the muscle’s maximum force-producing capacity (Powell et al. 1984). The most accurate measures of muscle volume are obtained using computed tomography or magnetic resonance imaging (Fukunaga et al. 1992; Elia et al. 2000; Eng et al. 2007; Barber et al. 2009), but recent work (Handsfield et al. 2014) created a number of regression equations to predict the volume of 35 muscles of the lower limb from the products of body mass and height. This method accounted for between 21 and 66% of individual plantarflexor muscle volume. Fukunaga et al. (1992) reported a muscle volume of 245 cm$^3$ and 140 cm$^3$ and mean PCSA of 68 cm$^2$ and 28 cm$^2$ for the medial and lateral heads of gastrocnemius respectively in 12 healthy adults using magnetic resonance imaging. Similar medial gastrocnemius volumes of $274 \pm 75$ mL and $303.2 \pm 65.1$ cm$^3$ were reported by Barber et al. (2009) and Morse et al. (2005a). Morse et al. reported lower physiological cross-sectional area values for both heads, of $48.5 \pm 6.4$ cm$^2$ for the medial and $25.6 \pm 4.9$ cm$^2$ for the lateral. Fibre lengths in the medial head were $4.5 \pm 0.23$ to $6.3 \pm 1.2$ cm and between $7.0 \pm$ and $7.4 \pm 0.34$ cm in the lateral (Maganaris et al. 1998; Morse et al. 2005a). Lengths have been shown to be greater in sprint athletes compared to both distance runners and controls (Abe et al. 2000). In females, fibre length may be up to 7% longer in the medial head and 2% longer in the lateral, with pennation (the angle the fibre makes with the aponeurosis/tendon) at the deep aponeurosis up to 6% greater in the medial head and 5% greater in the lateral (Chow et al. 2000). The range of pennation angle observed with change in joint angle was 20 to 52° in the medial head and 10 to 20° in the lateral. When contracted these ranges change to between 28 and 73° in the medial head and 16 and 41° in the lateral (Kawamaki et al. 1998).

Soleus is a broad flat pennate muscle composed primarily (>64%) of type I, slow-twitch fibres (Gollnick et al. 1973; Edgerton et al. 1975). It originates at the posterior head and upper quarter of the posterior fibular surface, the fibrous arch between the fibula and tibia and from the oblique line and the middle third of the medial tibial border. The vascular musculature lies between two aponeurotic lamellae. In normal, healthy soleus, the muscle runs downward becoming aponeurotic as it approaches the anterior gastrocnemius. The
fibres of the aponeurosis of the two muscles lie parallel to each other prior to merging. Soleus’ tendon then forms the deepest part of the Achilles (Drake et al. 2005). Abnormal soleus muscle may extend much further than normal, or as seen in ~2% of cases requiring Achilles surgery, may insert directly onto the upper surface of calcaneus without any tendon at all or via a separate tendon.

Mean volume of soleus has been shown to be 489 cm$^3$ and PCSA 230cm$^2$, approximately 3-8 times larger than the heads of gastrocnemius (Fukunaga et al. 1992). This is slightly smaller than the volume reported by Morse et al. of 520.1 ±114.0 cm$^3$ most likely due to body size differences. PCSA was smaller at 128.1 ±25.7 cm$^2$. Similar results in measures of posterior soleus fibre length and pennation angle were observed by both Morse et al. (2005a) and Maganaris et al. (1998), with fibre lengths of 4.3 ±0.8 cm and 3.5 ±0.4 cm and pennation angles of 25.3 ±3.2 and 25.0 ±2.6° reported by the two studies of male muscle architecture. Passive fibre length of between 2.4 and 4.8 cm, decreasing to between 2.1 and 3.6 cm when contracted was reported by Kawamaki et al. (1998) with values dependent on ankle joint angle. Pennation angle was between 16 and 28° during the passive condition and 30 to 52° when contracted. Fibre length has been shown to be between 3% less and 13% greater and pennation angle 8-27% less in females (Chow et al. 2000).

The Achilles tendon is the largest and strongest tendon in the body but is also the most frequently ruptured tendon in the lower limb, accounting for 20% of all large tendon injuries (McCory et al. 1999). At the junction with gastrocnemius and soleus, the Achilles tendon is a broad, flat tendon which becomes progressively more circular as it descends the limb, becoming more crescent-shaped approximately 4 cm from the osteotendinous junction with calcaneus (Doral et al. 2010). Bursae are present between the tendon and upper calcaneus and between the tendon and thickened deep fascia 5 cm above the muscle insertion to minimise friction (Sinnatamby 2006). The tendon fibres spiral as they descend the limb. Fibres originating from the tibial side of the muscle mass pass obliquely from deep to superficial aspect across the superficial surface of the tendon to the lateral side, while fibres from the fibular side pass on the deep side of the tendon to the tibial side of the tendon insertion. In most cases, the tendon twists through 90° but some double spiral, with the lateral head running from dorsal to ventral then back to dorsal (O'Brien 2005). Rotation begins before soleus merges, and if minimal fusion occurs the amount of rotation increases (Doral et al. 2010). The twisting has been suggested to contribute to the elastic properties of the tendon (Schepsis et al. 2002) but also increases the stress in the tendon.
which is most marked 2-5 cm above the insertion, an area which is poorly vascularised and one of the most common sites of Achilles tendinopathy and ruptures (Barfred 1971).

Tendon cross-sectional area refers to the amount of fibres arranged in parallel in the tendon and provides a measure of tendon force-producing capabilities. It can be determined using ultrasound and is typically reported in mm$^2$. Lichtwark and Wilson (2005b) reported a mean cross-sectional area of AT at the myotendinous junction of $53 \pm 3.6$ mm$^2$ (range: 48-59 mm$^2$) in a group of young adult males. Muraoka et al. (2004b) reported CSA of the AT the proximal, middle and distal tendon with significant differences noted between all three positions. Distal CSA was $64.7 \pm 9.9$ mm$^2$, at the middle it was $55.0 \pm 9.5$ mm$^2$ and at the proximal end it was $46.5 \pm 9.5$ mm$^2$. The minimum CSA was seen at 10% proximal to the insertion in 50% of participants and at 15% in the remainder, and was reported at $45.9 \pm 9.2$ mm$^2$.

2.2.2 The importance of the plantarflexors and plantarflexor assessment

The plantarflexors and plantarflexor assessment have been shown to be important in both dynamic performance and injury, with their main roles described below.

2.2.2.1 Locomotion

The plantarflexors have a vital role to play during locomotion, particularly during running. This is due in no small part to the stabilising and postural control function of soleus and the force production capabilities of gastrocnemius. Jacobs and van Ingen Schenau (1992) reported a normalised peak ankle moment of $3.15 \pm 0.10$ N.m.kg$^{-1}$ during the second stance of the sprint start in 7 good athletes (100 m personal best: $10.6 \pm 0.2$ s), similar to those observed by Arampatzis et al. (1999) during running at speeds between $2.61 \pm 0.18$ N.m.kg$^{-1}$ and $6.59 \pm 0.24$ N.m.kg$^{-1}$. Bezodis et al. (2014) analysed the joint kinetics of three elite athletes in the first stance phase, and similar to other authors found the primary role of the plantarflexors to be initially energy absorption and then force generation. Joint moments were normalised to both weight and leg length, and as a result the values are not comparable to other work with the exception of that of Bezodis et al. (2008) in the maximal velocity phase. The normalised moment in the first stance was much larger than that observed during maximal velocity, with observed values of 0.353 to 0.483 compared to 0.201 to 0.289 in the maximal velocity phase. These results support the importance of the plantarflexors in the acceleration phase.
Stefanyshyn and Nigg (1998) reported peak ankle moments of 300 N.m at the 15 m mark of the acceleration phase of the sprint, similar to the 300 N.m observed by Jacobs et al. (1993). Johnson and Buckley found the plantarflexor moment of 328 ±89 N.m was the largest moment (with the hip extensors) in stance during the mid-acceleration phase similar to Mann and Sprague (1980) during maximal velocity running. Kuitunen et al. (2002b) and Belli et al. (2002) reported slightly lower plantarflexor moments during maximal velocity sprinting of 240 and 233 ±29 N.m which may be due to the level of athletes used. Bezodis et al. (2008) reported larger moments of 336 ±83 N.m in a group of four elite athletes measured at 45 m of the sprint supporting the findings of Hobara et al. (2008) who found trained sprinters generate larger moments with a stiffer ankle.

Analysis of joint work and power has shown the primary role of the plantarflexors to be force absorption in initial stance and force generation in late stance, but caution is advised in interpretation of joint powers due to reported differences in underlying MTU behaviour in terms of force absorption and generation (Cronin et al. 2013). Increased use of ultrasound during dynamic tasks such as sprinting provides further insight into plantarflexor MTU behaviour during dynamic tasks.

Electromyography (EMG) analysis lends further support and insight into the importance of the plantarflexors in locomotion. The plantarflexors have been shown to be highly active immediately prior to ground contact, during the braking phase of contact and immediately prior to toe-off in running, typically increasing linearly with speed. Increased plantar- and dorsiflexion muscle pre-activation is important in the regulation of landing stiffness (Gollhofer and Kyrolainen 1991) as it acts to increase ankle stiffness and improve ability to apply force rapidly. Pre-activation may also increase muscle spindle sensitivity via α-γ-coactivation potentiating stretch reflexes (Gottlieb et al. 1981) which enhance musculotendinous stiffness (Nichols and Houk 1976; Hoffer and Andreassen 1981). This increased stiffness prevents unnecessary yielding during the braking phase of ground contact (Kyröläinen et al. 2001). Pre-activation may also allow for the tendon to work in its elastic region during ground contact, where it is stiffer than in the toe region. This increase in stiffness may be desirable as it decreases the ability of the muscle to shorten rapidly which subsequently decreases metabolic demand (Lichtwark and Wilson 2005a).

At slow speeds (up to 1.75 m.s⁻¹) both soleus and gastrocnemius activity has been shown to increase with increased speed (Hof et al. 2002a). Increases were also observed at higher
speeds (2.25 to 4.5 m.s\(^{-1}\)), with increased soleus activation by \(\sim\) 20\% to 120 μV and a 40\% increase in gastrocnemius medialis and lateralis activity (Gazendam and Hof 2007). Kuutunen et al. (2002b) reported a \(\sim\) 20\% increase in soleus activity in athletes running between 70 and 100\% of maximum speed but only a 20 μV (\(\sim\) 13\%) change in gastrocnemius activity. Kyröläinen et al. (1999) found gastrocnemius EMG increased from 200 μV in the 100 ms prior to touchdown during running at 3.25 m.s\(^{-1}\) to between 400 and 500 μV during maximal speed running (\(\sim\) 8.3 m.s\(^{-1}\)). During stance, activity increased to \(\sim\) 600 μV during maximal sprinting compared to \(\sim\) 400 μV at slower speeds. Similar increases in gastrocnemius EMG were observed by Kyröläinen et al. (2005), with EMG of \(\sim\) 400 μV observed during running at 4.0 m.s\(^{-1}\) which increased to between 800 and 1000 μV during maximal speed running (8.5 ±0.57 m.s\(^{-1}\)). One of the main benefits of this increase in muscle activity is an increase in muscle stiffness, which aids movement efficiency and aids in reducing ground contact time. This was reported by Chumanov et al. (2012), who found that when step rate was modified from preferred to 110\% of step rate, gastrocnemius activity significantly increased supporting its role in determination of contact time and stiffness modulation.

The plantarflexors have also been proposed to influence the timing of the walk-run transition, with one proposed reason a need for the body to keep the muscle group operating at its optimal force production level. The dorsiflexors have been proposed to reach an optimal level of activation with an increase in angular velocity, beyond which they believe they are overexerting (Hreljac et al. 2001) which then triggers a transition into a run. Neptune and Sasaki (2005) proposed that a similar mechanism in the plantarflexors also triggers the transition. Increased walking speed shifts the muscle into an area of its force-velocity curve not conducive to force production, but a transition into a run moves the muscle shortening velocity back into the optimal area of the curve. This was observed \textit{in vivo} by Farris and Sawicki (2012b). Bartlett and Kram (2008) lend further support to this theory of transition as they found reducing plantarflexor EMG increased the walk-run transition point to 2.06 ±0.04 m.s\(^{-1}\) whereas increased activity reduced the transition point to 1.80 ±0.03 m.s\(^{-1}\). However, no summation effect was observed and EMG in those muscles also continued to increase with the increase in speed (as observed by several other authors) which suggests that something other than just these factors is affecting the transition point (Bartlett and Kram 2008).
Other authors have observed the body will always try to maintain movement efficiency, which may be achieved by shifting force generation to muscle groups better anatomically suited to optimise movement efficiency. During running, there is a greater utilisation of the SE portion of the MTU (Sasaki and Neptune 2006b; 2006a). Farris and Sawicki (2012a) found there was no difference in the proportion of power contributed by the hip, knee and ankle joints to total power across speeds within a gait, but there was a significant shift in power production from the hip to the ankle with the change in gait, which may explain observed increased efficiency at speeds above 2.0 ms\(^{-1}\). Gastrocnemius and soleus are designed for efficient energy storage and utilisation due to their long Achilles tendon and short muscle fibres (Lichtwark and Wilson 2008).

2.2.2.2 The role of the ankle in leg stiffness modulation

Stiffness refers to the ratio of applied force to the amount of deformation of a body (Zatsiorsky and Prilutsky 2012). The simplest model of stiffness control in the human body is the spring-mass model, where the body is represented as a solid mass on top of a compressible spring (Butler et al. 2003). Leg and vertical stiffness are commonly reported stiffness measures in sports biomechanics literature, with high reliability observed for the measure of leg stiffness (McLachlan et al. 2006a). Latash and Zatsiorky (1993) refer to these measures as quasi-stiffness, i.e. composite stiffness which reflects the stiffness of all the joints and component tissues (such as muscle, tendon and ligament) of the lower limb. Due to their location and size, the plantarflexors have an important role to play in modulation of ankle joint stiffness which has a subsequent influence on leg stiffness.

Measures of vertical stiffness while hopping on the ground and jumping on a sledge have been shown to be highly reliable (Flanagan and Harrison 2007; Joseph et al. 2013). Joseph et al. (2013) reported good reliability for vertical stiffness during running and hopping at 2.2 Hz with one or both legs. Good reliability was defined as a mean difference of less than 5%, ICC ≥ 0.80, coefficient of variation of less than 10% and a between-trials effect size of less than 0.60. Hopping at a self-selected frequency was much less reliable, with only one of these criteria met. Similarly, measures of knee and ankle stiffness were also very unreliable. Flanagan et al. reported very high ICCs for stiffness and force using the lower body sledge. Single ICC was greater than 0.863 for the drop jump and 0.907 for rebound jumps, with average ICC >0.950 and 0.967 for the two protocols. Differences in results are probably
due to the increased control of the sledge apparatus. No published study appears to have used a force sledge to determine joint stiffness.

Increased running speed is associated with increased leg stiffness, but not necessarily ankle joint stiffness. Stefanyshyn and Nigg (1998) reported an increase in ankle stiffness was associated with an increase in running speed, but this may be because the participants at higher speeds were trained sprinters who are known to demonstrate higher joint stiffness during dynamic tasks (Hobara et al. 2008). Other literature shows no change in ankle joint stiffness with increased running speed (Arampatzis et al. 1999; Kuitunen et al. 2002b). The knee appears to be a more important modulator of leg stiffness in running. This is probably due to the increased size of the quadriceps and hamstring muscles relative to the ankle plantar- and dorsiflexors, as increased muscle size (hence increased area for muscle activation) has been shown to be associated with increased stiffness (Ryan et al. 2009).

Ankle stiffness during hopping is also of interest to researchers (Farley and Morgenroth 1999; Hobara et al. 2009; Hobara et al. 2011; Kuitunen et al. 2011). Magnitude of ankle stiffness has been shown to remain relatively constant with increased demand for decreased ground contact time (Kuitunen et al. 2011) despite large increases in medial gastrocnemius and soleus activation during the first 50 ms after landing when there is a need for decreased contact time (Hobara et al. 2007). During hopping at 2.2 Hz and 3.0 Hz, it has been shown to be the main determinant of leg stiffness (Hobara et al. 2011). At 1.5 Hz, the knee is the primary determinant of stiffness as the large muscle mass is advantageous in development of a large moment during a maximal task (Hobara et al. 2009).

2.2.2.3 Injury

Recreational running has become increasingly popular in recent years but this has also led to an increasing amount of injuries presenting to healthcare professionals. Approximately 50% of runners will experience injury each year, with 25% injured at any given time (Fields et al. 2010). The calf and ankle region is the second most injured area of the lower limb, after the knee. Dependent on injury definition, percentage of running injuries occurring in the calf and ankle area ranges from approximately 25.9% (Taunton et al. 2002) to 57.7% (Maughan and Miller 1983; van Middelkoop et al. 2008; Buist et al. 2009; Chang et al. 2013). Fong et al. (2007) found the ankle to be the most common site of injury in 24 out of 70 sports, most of which required high levels of stretch-shortening cycle function due to a need for elastic propulsion or impact absorption, or increased ankle stability. The incidence
of injury in other sports varies from 14.5% in rugby league sevens (King et al. 2006) and 16.2% in Australian Football (Gabbe and Finch 2001) up to 45.7% in artistic gymnastics (Kirialanis et al. 2003), 56.8% in soccer refereeing (Bizzini et al. 2009) and 66% in netball (McManus et al. 2006).

Overuse injury in the calf muscles and Achilles tendon is common, due to their importance in locomotion. This is a generic term used to describe a clinical condition of a tendon due to overuse characterised by a combination of pain, swelling (diffuse or localised) and impaired performance (Maffulli et al. 1998). While high levels of repetitive physical activity and sudden increased running mileage are established risk factors of developing a tendinopathy, there are numerous other risk factors that predispose people to the condition such as decrease in recovery time, change in training surface, incorrect/excessively worn footwear, excessive hill work, lack of correct conditioning, obesity, hypertension, high cholesterol, use of steroids and oestrogen compounds, muscle weaknesses and imbalances, anatomical malalignment, weak plantarflexor muscles, increased internal tendon temperature, increased age, height and weight (Kvist 1994; Kader et al. 2002; Maffulli and Kader 2002; Maganaris et al. 2004a; Fredberg and Stengaard-Pedersen 2008; Farris et al. 2011). As a result, even sedentary, non-athletic populations have been known to be affected by tendinopathy (Sayana and Maffulli 2007).

There appears to be a trend of altered stiffness in those presenting with lower limb overuse injury. Increased leg stiffness is typically associated with improved performance in stretch-shortening cycle (SSC) tasks but there appears to be an optimal level of stiffness which balances performance enhancement and injury risk. Williams et al. (2001; 2004) found increased knee stiffness in runners with high arches which was associated with an increased risk of ankle injury. In contrast, those with low arches tended to run with a stiffer ankle and were more likely to suffer from a knee injury. The tendency for an increased risk of injury to the area with decreased stiffness has also been observed in patients with Achilles tendinopathy. Similar to the findings of Helland et al. (2013) in patients with patellar tendinopathy, Arya and Kulig (2010) found patients with Achilles tendinopathy had lower tendon stiffness than healthy controls by approximately 25% (375.25 ±61.88 N.mm$^{-1}$ vs. 300.37 ±37.60 N.mm$^{-1}$). This was due to increased elongation despite decreased force exerted on the tendon. Young’s modulus in the tendinopathic patients was less than half that of the healthy participants (1,671.02 ±277.50 MPa vs. 818.72 ±217.03 MPa). A similar decrease in Achilles tendon stiffness in the presence of tendinopathy was observed by
Child et al. (2009). Lowered stiffness may result in increased repetitive excessive strain of the tendon and subsequent injury. These observations may help explain reported lower stiffness in a limb affected by Achilles tendinopathy compared to its ‘healthy’ contra-lateral control in 84% of participants during a jumping task observed by Maquirriain (2012).

2.2.2.4 Anterior cruciate ligament injury

Due to their anatomical location, the plantarflexor muscles have been suggested to be of importance in understanding the mechanisms of anterior cruciate ligament (ACL) injury (Mokhtarzadeh et al. 2013). Gastrocnemius is usually considered the more important muscle due to its insertion on the distal femur, but soleus also has a role to play due to its insertion on the posterior tibia and fibula. The effect of gastrocnemius on the ACL was originally proposed by O’Connor (1993) using a simulated model who found gastrocnemius loads the ACL throughout the full range of knee flexion. Fleming et al. (2001) reported similar results during in vivo loading of gastrocnemius using a twitch technique, with the magnitude of strain dependent on ankle torque and knee flexion angle. At 5° and 15° of flexion strain of up to 3.5% was observed with negligible strain observed at higher angles. Increased strain was associated with increased ankle torque, however, Shimokochi et al. (2009) found that increased ankle and hip torques during landing due to participants slightly leaning forward greatly reduced knee extensor torque. This suggests that the results of isolated analyses of strain during controlled tasks must be considered with inverse dynamics analyses of real-life landing activities.

The timing of muscle activation and force production must also be considered in determining the role of the plantarflexors in ACL injury. Elias et al. (2003) found soleus activation resulted in posterior tibial translation, hence the earlier activation of soleus compared to gastrocnemius in a landing task may act to protect the ligament. Simultaneous activation of soleus and gastrocnemius resulted in anterior tibial translation and increased ACL strain, in contrast to the findings of Sherbondy et al. (2003). They found increased tension in the plantarflexors resulted in posterior tibial translation. The in vitro nature of the testing of Elias et al. (2003) with elderly male cadaver knees must be considered as they do not necessarily replicate that of young females who are the primary group affected by this injury. Differences in relative muscle mass and activation potential must also be considered, as soleus has a physiological cross-sectional area of between 3 and 8 times greater than that of gastrocnemius which may explain why Sherbondy et al. (2003) observed the opposite
effect to Elias et al. (2003) in intact, healthy human limbs. Mokhtarzadeh et al. (2013) further supported the importance of the plantarflexors in landing with reported large soleus forces in late landing (30% of that of the hamstrings) and low gastrocnemius forces in early landing.

As ACL injuries are most likely to occur 17 to 50 ms post-landing (Krosshaug et al. 2007), a decreased plantarflexor force production capability in this period of time could place athletes in a potentially risky situation during landing. The demand placed on the plantarflexors by acting as stiffness modulators during plyometric activities may also partly explain why these exercises are effective in reducing risk of ACL injury (Hewett et al. 2006a; Myer et al. 2006).

2.2.2.5 Section summary

The plantarflexor muscles have an important role to play in locomotion, and are frequently injured. While assessment of the overall MTU is important, it is also important to be able to accurately assess the function of the individual components of the MTU. The next section will provide an overview of the main measures of interest for the muscle and tendon of this muscle group, provide some normative data and identify the main reasons why variations in value may be observed.

2.3 Measures of plantarflexor function of interest

Section 2.2 outlined the importance of the plantarflexors in everyday tasks, but to understand how they contribute to each role they play requires assessment of a range of physiological and mechanical measures. The following section provides a synopsis of the most relevant measures of plantarflexor MTU function and normative values for each.

2.3.1 Force and force development

The question of whether stride length or stride frequency determines running speed has fascinated researchers for years. The deterministic model shows how different factors determine these two variables (Figure 5), but these can all be influenced by plantarflexor performance in a SSC task.
The running literature suggests that up to running speeds of 7 m.s\(^{-1}\) force generation is a key determinant of running speed (Weyand et al. 2000; Mercer et al. 2002; Mercer et al. 2005). This is a result of the increased ground contact time, which allows for development of increased force. In contrast, at speeds above 7 m.s\(^{-1}\), athletes will increase their stride frequency which can be achieved by modulating the limb stiffness response, to compensate for the decreased time to develop force. These variables, their assessment and enhancement has been the focus of a vast quantity of research over the years (Alexander 1989; Farley and Gonzalez 1996; Weyand et al. 2000; Belli et al. 2002; Bret et al. 2002; Günther and Blickhan 2002; Kuitunen et al. 2002b; Hunter et al. 2004a; Hunter et al. 2004b; Hunter et al. 2005). The contribution of the plantarflexors to stiffness and speed are described in section 2.2.2.3 therefore will not be repeated here. The focus of this section is on plantarflexor force and force development measurement.

Dynamometry is the most commonly used method of assessing force production capabilities due to its capability to isolate the muscle group and the level of control this method affords (Baltzopoulos et al. 2012). Both isometric and isokinetic plantarflexor torque have been shown to be influenced by a number of factors, including gender, age, training and dominance. Isometric torques in young adults range from 103-104 N.m (Kubo et al. 2005b) and 115.8 ±25.2 N.m (Kubo et al. 2007a) up to 157.2 ±28.3 N.m in young male distance runners (Fletcher et al. 2010). Pääsuke et al. (2000) reported large, statistically significant age-related declines in isometric MVCs from the 3\(^{rd}\) decade onwards. Values decreased from 842.2 ±42.1 N and 817.9 ±34.8 N in the 3\(^{rd}\) and 4\(^{th}\) decades to 489.0 ±41.3 N in the 8\(^{th}\) decades, a decrease of 72%. Accounting for a 16-18 cm lever length, the equivalent torques dropped from between 134-151 N.m and 78-88 N.m.
Mademli and Arampatzis (2008) reported plantarflexor torque of 85.9 ±17.9 N.m in 14 elderly males which is greater than the 63.5 ±21.1 reported by Kubo et al. (2007). These values were significantly lower than the torque of younger participants (115.8 ±25.2 N.m). Differences were still significant even when normalised to body mass (young: 1.62 ±0.30, old: 1.06 ±0.30) and muscle thickness (1.55 ±0.31 and 1.00 ±0.28). The main reason proposed for the decrease in plantarflexor torque is decreased muscle mass, but these results would suggest other factors such as neural control may also influence output.

In healthy young adults, values reported for plantarflexor concentric torque in the dominant limb range from 71.3 to 184 N.m at 30 deg.s\(^{-1}\) and from 84.3 ±14.6 to 145.4 N.m at 60 deg.s\(^{-1}\) (Falkel 1978; So et al. 1994). At an angular velocity of 45 deg.s\(^{-1}\), peak concentric torque in 37 young adult males was 86.3 ±26.8 N.m (Poulis et al. 2000). Differences in magnitude are primarily related to training background but may also due to differences in the protocol used for setup, with knee angle known to influence plantarflexor torque due to the biarticular nature of gastrocnemius. Testing and analysis protocols used may further influence observed results (Baltzopoulos et al. 2012).

A muscle group needs to be able to generate large amounts of force, but also apply them rapidly. The rate of force development of a muscle group has previously been linked to the stiffness of the muscle’s tendon (Burgess et al. 2007). Cannavan et al. (2012) reported normalised rate of force development of the plantarflexors in the first 50 ms of isometric contraction with the knee at 180° to be between 4.3 ±4.6 N.m.s\(^{-1}\).kg\(^{-1}\) and 4.6 ±4.2 N.m.s\(^{-1}\).kg\(^{-1}\), slightly lower than values observed in the knee extensors. These values are slightly greater than the 4.12 ±1.89 N.m.s\(^{-1}\).kg\(^{-1}\) observed in healthy older adults by LaRoche et al. (2010) with the knee at 105°. Thelen et al. (1996) reported a much greater maximum rate of torque development in the plantarflexors of 608 ±169 N.m.s\(^{-1}\) in young females, 957 ±248 N.m.s\(^{-1}\) in young males, 389 ±171 N.m.s\(^{-1}\) in older females and 681 ±223 N.m.s\(^{-1}\) in older males, with significant effects of age observed. The maximum rate of force development is related to the maximum force generated, and values of 130 ±27 N.m, 181 ±38 N.m, 88 ±21 N.m and 137 ±38 N.m were reported for each group respectively. This decreased ability to generate force rapidly has been suggested by a number of authors to be related to decreased stability during dynamic tasks. Finni et al. (1998) reported a rate of force development in the Achilles tendon during walking at 1.1 m.s\(^{-1}\) of 6570 ±1810 N.s\(^{-1}\), 8430 ±3090 N.s\(^{-1}\) during walking at 1.5 m.s\(^{-1}\) and 9670 ±3260 N.s\(^{-1}\) during walking at 1.8 m.s\(^{-1}\). This large range of values was not associated with a similarly
large range of vertical force values, which suggests that observed external forces may not necessarily reflect internal muscle and tendon forces (Fukashiro et al. 1993).

2.3.2 Stretch-shortening cycle function

In everyday tasks such as running, hopping or throwing, muscular contraction rarely occurs using just an eccentric, concentric or isometric contraction (Komi 2000). Instead, the MTU lengthens in an activated condition and shortens. This is referred to as a SSC, where an eccentric contraction is immediately preceded by a concentric contraction resulting in performance enhancement in the concentric phase (Bosco and Komi 1979). The effectiveness of a person’s ability to utilise the SSC, i.e. their ability to effectively utilise eccentric loading, is usually assessed using the reactive strength index (RSI) during drop jumps or cyclical loading. This refers to the ratio between a jump height and its preceding ground contact time (Young 1995).

One of the underlying factors affecting an athlete’s ability to utilise the SSC is the mechanical properties of their MTU (Anderson 1996). Athletes with increased MTU stiffness have been shown to have enhanced ability to utilise the SSC (Butler et al. 2003; Harrison et al. 2004) most likely due to increased energy storage ability in the MTU and increased reflex activity during lengthening. SSC effectiveness is also highly related to the characteristics of the prior loading which ideally should be analysed on an individual muscle basis as composite MTU behaviour or joint power analysis does not represent individual muscle or tendon actions (Sousa et al. 2007; Cronin et al. 2013; Earp et al. 2014). Due to their excellent ability to control eccentric loading, force sledges have been used with ultrasound to conduct a number of studies into the quadriceps and plantarflexor muscle response to loading (Ishikawa et al. 2003; Ishikawa and Komi 2003; Ishikawa et al. 2006a; Hoffren et al. 2007; Sousa et al. 2007). Increasing loading (by increasing drop jump height) typically enhances concentric performance as assessed by rebound velocity but only to a point, beyond which no gains are to be achieved but which may potentially be injurious. Increased loading, similar to that observed in increasing speed, increased soleus and gastrocnemius activity particularly in the push-off phase. Individual muscle differences varied with height (Sousa et al. 2007). The magnitude of muscular strain was muscle-dependent, similar to results observed during a walking task (Ishikawa et al. 2005). Earp et al. (2014) showed in an isolated knee flexion SSC task that when muscle strain does not change, tendinous strain significantly does as suggested by its viscoelastic nature. At lighter
loads it tends to act as a power amplifier which increases the relative duration of the movement, whereas at heavier loadings it tends to act more as a rigid force transducer allowing transfer of large muscular forces.

![Figure 6. The stretch-shortening cycle showing pre-activation which aids impact absorption (A), stretch/eccentric phase (B) and shortening/concentric phase (C) (from Komi, 2000)](image)

There are three main proposed mechanisms as to why the stretch of the MTU is effective in enhancing subsequent concentric performance. There is considerable debate about which mechanism explains the force enhancement, but review of the literature suggests that the mechanism of effectiveness is almost certainly task-dependent (Wilson and Flanagan 2008).

1. **It allows muscle time to develop force**

   Even when instructed to perform an action as quickly as possible, it still takes time to develop force in the muscle. The length of time is dependent on how long it takes the central nervous system to generate the signal, length of time required for excitation-contraction coupling to occur as well as the mechanical properties of the MTU (Cavanagh and Komi 1979). MTU work is submaximal following immediate concentric contraction with increased force as the shortening distance of the MTU is moving at submaximal force. Inclusion of an eccentric phase of contraction allows for increased time for force to develop, as there is increased time for cross-bridge activation to occur. Further support for this mechanism was suggested to be the enhanced performance observed in slow-twitch compared to fast twitch muscle during a longer duration SSC (van Ingen Schenau et al. 1997a). This mechanism
cannot explain performance in rapid SSC tasks such as sprinting and bounding, where there is only very short time (100-150 ms) to develop force.

2. It allows for storage and utilisation of elastic energy

There is no question that inclusion of an eccentric contraction allows for storage and reutilisation of elastic energy within the MTU (van Ingen Schenau et al. 1997b), but whether this mechanism can contribute to the enhanced force output during the concentric phase is in debate (Wilson and Flanagan 2008). Elongation of the muscular and tendinous structures of the limb allows for storage of elastic energy in the SEE due to the law of the conservation of energy, but how effective this storage is appears to be highly dependent on the duration of the loading and time taken to transition to the concentric phase. Due to the need for a very short amortisation phase to allow for reutilisation of the stored elastic energy, this mechanism is more likely to explain effectiveness in activities with high, rapid force requirements such as sprinting rather than the countermovement jump.

3. There is a contribution to output from reflexes

Komi (2000) proposed a strong argument for the potential role of stretch reflexes in SSC effectiveness. As outlined in 2.2.2.1, there is a very high level of muscle pre-activation immediately prior to ground contact in running and in the early stages of contact (braking phase). The subsequent muscle elongation is then very short and fast, with an almost immediate transition to the concentric phase. These conditions may combine to trigger stretch reflexes within the MTU which act to enhance the subsequent concentric phase performance (Komi 2000) but this is heavily dependent on a high stretch velocity in the MTU. During fast stretch-shortening cycles (<0.25 s ground contact time) the rate of loading may be sufficient to trigger a reflex response, but it is unlikely to occur during a task such as a countermovement jump where there is a relatively slow rate of loading and potentially long amortisation phase. The presence of increased muscle activity would support the role of reflexes to enhance SSC effectiveness, but this was not observed by Finni et al. (2000) during a countermovement jump. This lends further support to the argument that different mechanisms may explain force enhancements in tasks with different loading requirements.
2.3.3 Muscle-tendon unit mechanical behaviour

Tendon can be flat or round and is located at the middle, origin or insertion of a muscle. It is characterised by 6 key mechanical features:

1. Tangent modulus increases in a non-linear fashion at the beginning of loading, followed by a linear portion
2. Hysteresis occurs in the loading-unloading cycle
3. Under constant load significant creep occurs; under constant elongation significant stress relaxation occurs
4. The stress-strain curve is dependent on the rate of deformation
5. Instantaneous properties are dependent on the deformation, but with preconditioning these properties become constant
6. During stretch, the collagen fibres straighten (Fung 1981; Ault and Hoffman 1992).

There are numerous MTU properties reported in the literature from both in vivo and in vitro studies, ranging from the cross-sectional area to the ultimate failure point. This section of the literature review will provide a brief introduction to the main mechanical properties of interest from in vivo studies.

2.2.3.1 Stress

Tendon stress refers to the ratio of force in the tendon to tendon cross-sectional area (CSA) (Burgess et al. 2009b) i.e.

$$\sigma = \frac{F}{T_{CSA}}$$

where $\sigma$ = stress, $F$ = force generated during contraction and $T_{CSA}$ = tendon cross-sectional area (Buchanan and Marsh 2001). The measurement of tendon force highly affects the measure of stress, and reported values range from 30.28 ±3.73 MPa and 34.50 ±4.45 MPa in young adult females (Burgess et al. 2009c), up to 80 ±21 MPa in young adult males (Peltonen et al. 2010) and 81.2 ±17.0 MPa in young adult males and females (Waugh et al. 2012).
2.2.3.2 Strain

Tendon strain refers to the amount of tendon lengthening per unit resting tendon length, i.e.

\[ \varepsilon = \frac{dL}{T_L} \]

where \( \varepsilon \) = strain, \( dL \) = tendon elongation and \( T_L \) = resting tendon length (Buchanan and Marsh 2001). One of the main problems with strain calculations is the determination of resting tendon length, which can be difficult due to the presence of the toe region (Ker 2007). Some researchers have taken tendon resting length as the length of the tendon at a certain loading e.g. 200 N (Lichtwark and Wilson 2005b) and others have taken it as the length during upright standing (Farris et al. 2012). Measures of peak strain tend to be quite variable, influenced by both the measure of resting length as well as being influenced by the magnitude of loading. Kubo et al. (2007a) reported Achilles tendon strain of 5.0 ±1.3% in young adults and strain of 3.8 ±1.0% in older adults. Waugh et al. (2012) reported strain of 9.2 ±2.2% during isometric loading which is greater than the 6.2 ±2.6% reported by Peltonen et al. (2010) during the same protocol, who also reported a large range of values from 4.1 to 12.4%. The values reported by Waugh et al. are similar to those of Lichtwark and Wilson (2005b) during one legged hopping (mean strain: 8.3 ±2.1%; range: 4.3-11.4%). Using the same protocol as Lichtwark, Farris et al. (2012) found much lower values for strain of 3.5 ±1.8 and 3.8 ±2.0% which they attributed to the difference in determination of resting tendon length.

2.2.3.3 Young’s modulus

Young’s modulus refers to the ratio of linear stress to uniaxial strain in the range of stress in which Hooke’s Law applies, and is a measure of stiffness normalised to tendon dimensions. Burgess et al. (2009b) calculated Young’s Modulus as:

\[ E = k \times \frac{T_L}{T_{CSA}} \]

where \( E \) = Young’s modulus, \( k \) = stiffness, \( T_L \) = resting tendon length and \( T_{CSA} \) = tendon cross-sectional area. Lichtwark et al. (2005b) defined elastic modulus as the slope of the mean stress-strain curve for each participant, with typical values ranging from 0.61 GPa to 1.15 GPa (mean ± S.D: 0.87±0.2). Similar values of between 0.60 ±0.05 and 0.64 ±0.07
GPa were reported by Burgess et al. (2009c) in young adult females with changes in hormone level across the menstrual cycle, but much lower values of 0.32 ±0.02 and 0.36 ±0.04 GPa were reported in elderly males and females. Peltonen et al. (2010) reported a Young’s modulus of 2.0 ±1.0 GPa whereas his later work reported a much lower value of 0.78 ±0.33 GPa (Peltonen et al. 2012).

2.2.3.4 Creep

Creep is defined as the characteristic of viscoelastic materials where it undergoes increased deformation under a constant stress until an asymptotic level of strain is reached (Pearson et al. 2007). It is calculated as the distance between the start point and end point of each loading-unloading cycle (Legerlotz et al. 2007). The presence of creep has been reported by a number of authors during isometric testing, with most authors using a pre-conditioning protocol involving a number of isometric contractions before testing to account for this property. Maganaris (2003b) suggested that steady-state behaviour occurred after just five three-second loading and one-second maintenance isometric contractions, but Hawkins et al. (2009) recommended 270 loading cycles or a six minute warm-up to ensure steady-state mechanical behaviour as dynamic creep of between 1.64% to 5.94% was observed in participants. This property does not appear to be as relevant in studies involving repetitive SSC. Farris et al. (2012) reported no change in tendon length from resting levels even after a 30 minute run, suggesting that the short duration contact times in running (<0.3 s) are not sufficiently long enough to induce this effect in tendon and the cyclical rather than constant loading allows tendon to recover in between. They reported a tendon length of 301 ±16 mm before a 30 minute run and a length of 300 ±14 mm afterwards. Pearson et al. (2007) had reported a similar time-dependency of the magnitude of creep, as they found much lower creep and strain, and higher stiffness during short-duration (~3s) compared to longer-duration (~10s) contractions. To allow for comparison of stiffness and strain measures, they emphasised the importance of standardising loading duration.

2.2.3.5 Hysteresis

Tendon is an elastic material therefore typically exhibits hysteresis, which provides a measure of energy converted to heat. A commonly definition is the area under the ascending force-elongation curve minus the area under the descending curve expressed as a percentage of the area under the ascending curve (Lichtwark and Wilson 2005b; Burgess et
Lichtwark and Wilson (2005b) reported hysteresis of the Achilles tendon of 26 ±9.2% during single-legged hopping (interquartile range of 17-35%).

2.2.3.6 Tendon stiffness

Stiffness refers to the ratio between an applied force and the deformation of a body (Zatsiorsky and Prilutsky 2012), and can be applied to every aspect of the human body from the muscle fibril to the whole body (Butler et al. 2003). Stiffness of the leg, joint and musculotendinous structures is thought to have a major influence on the rate of force development in the lower limb, sprint kinematics and elastic energy storage and utilisation (Brughelli and Cronin 2008b; 2008a). The formula used to calculate stiffness is dependent on the structure in question. The contribution of ankle stiffness to leg stiffness has been discussed previously in 2.2.2.2, and is potentially related to the underlying stiffness of the muscle and tendon. This section focuses on measures of Achilles tendon stiffness.

A range of methods are used in practice which can have large effects on computed values (Pearson and Onambélé 2012). Methods include determining the slope of a linear polynomial fitted to a section of the force-elongation curve, typically at levels greater than 50 or 60% of maximum force achieved, as this eliminates the toe region of the curve (Kubo et al. 2007b; Farris et al. 2012). Lichtwark and Wilson (2005b) calculated stiffness as the slope of the entire force-elongation curve as they assumed by only analysing forces over 200 N the toe region was already eliminated. Stiffness can also be calculated as the slope of the tangent at a particular point on the curve such as 90 or 100% of maximum force, which makes comparison of results quite difficult. Pearson and Onambélé (2012) showed that analysing a curve from 0-100% could produce a stiffness value of less than half that observed as the tangent at 100% of maximum force, and approximately 50% less than that observed as the slope of the curve from 50% to 100% of maximum force. When a ‘reference standard’ stiffness (i.e. average stiffness at every 10% of the force-elongation curve was compared to each of the other methods, adjustment ratios of between 0.74 (for the entire slope-elongation curve) and 1.51 (for the tangent at 100% MVC) were obtained.

This difference in method of calculation is the main source of difference between observed stiffness values in the literature. Burgess et al. (2007) reported values of between 43.9 ±2.5 N.mm\(^{-1}\) and 49.0 ±10.8 N.mm\(^{-1}\) for tendon stiffness prior to an endurance and plyometric training intervention which increased stiffness to 71.0 ±7.4 N.mm\(^{-1}\) and 63.4 ±9.2 N.mm\(^{-1}\). Kubo et al. (2007b) reported stiffness of 129.0 ±35.8 N.mm\(^{-1}\) and
154.0 ±55.2 N.mm$^{-1}$ before and after a similar plyometric training programme. Arampatzis et al. (2006) found stiffness values to be much lower, between 20.6 ±6.6 N.mm$^{-1}$ and 34.6 ±6.6 N.mm$^{-1}$ in a group of trained runners, values only slightly greater than those observed by Kubo et al. (2008) in elderly walkers (between 12.7 ±8.4 N.mm$^{-1}$ and 14.1 ±8.9 N.mm$^{-1}$). Waugh et al. (2012) found stiffness in young adults to be 237.6 ±78.9 N.mm$^{-1}$ which is in a similar range to that observed by Peltonen et al. (2012) (197 ±62 N.mm$^{-1}$ to 213 ±56 N.mm$^{-1}$). Slightly lower stiffness values of 188 ±43.2 N.mm$^{-1}$ (range: 125-234 N.mm$^{-1}$) and 147 ±52 N.mm$^{-1}$ to 163 ±41 N.mm$^{-1}$ were reported by Lichtwark and Wilson (2005b) and Farris et al. (2012).

2.3.4 Section summary

This section has presented a number of muscle-tendon unit function variables that may be of interest to the researcher, and presented a brief introduction to the stretch-shortening cycle and its proposed mechanisms of effectiveness.

2.4 Assessment of plantarflexor MTU function

The literature described above highlights the importance of the plantarflexors, and some potential variables of interest. The following section will present a synopsis of methods of investigating and assessing MTU function, with a particular emphasis on voluntary, active measurements. Animal models will be briefly discussed due to their importance to the literature on the MTU response to intervention.

2.4.1 Animal models of function

Research with animal subjects allows study of the MTU response to loading in highly controlled conditions using methods that are not feasible in humans. These measures are usually highly invasive e.g. sonomicrometry (Buchanan and Marsh 2001), force buckle transducers in the tendon (Biewener and Baudinette 1995) or removal of tendon from the test animal and clamping of the tendon into a test machine (Landis et al. 1995; Almeida-Silveira et al. 2000; Kuo et al. 2001; Legerlotz et al. 2007). Research in animal MTU can be conducted in vivo, ex vivo or in vitro, the latter two differentiated by the length of time the tendon is outside the body (greater or less than 24 hours). Research questions in animal studies range from determination of the ultimate failure point of tendon (Soslowsky et al. 2000; Devkota 2003; Huang et al. 2004), biochemical and histological analysis (Landis et al.
1995) and effects of pre-conditioning and denervation (Almeida-Silveira et al. 2000; Sverdlik and Lanir 2002; Arruda et al. 2006; Trudel et al. 2007) to responses to continuous sinusoidal loading (Ker 1981), direct measurement of force during dynamic movements (Biewener and Baudinette 1995; Meyer et al. 2004; McGuigan et al. 2009) and response to training methods (Woo et al. 1980; Legerlotz et al. 2007). They have also been used to investigate injury mechanisms, with Salowsky et al. (2000), Huang et al. (2004) and Gimbel et al. (2004) using animal models to investigate injury in the supraspinatus, Achilles tendon and rotator cuff muscles respectively. The area approximately 6 cm superior to the calcaneal insertion of the Achilles tendon is one of the most common sites of Achilles tendinopathy, and Landis et al. (1995) found points of increased stress in the tendon demonstrated increased calcification which may explain this increased tendency for injury.

Animal-based research is invaluable in informing us about MTU behaviour and the mechanical response to loading. However, many question the ethics of this type of research especially when there are obvious differences in the mammals from which the test materials are obtained compared to humans. Similarities between animal and human tendon even within the same tendon are unclear due to differences in lever lengths, loadings and potentially rates of collagen synthesis and circulating hormones. In vitro tendon behaviour may not reflect that which occurs in vivo, and treatment and fixation methods used may negatively influence observed in vitro results. This limits the applicability of animal models of research to study MTU responses to loading in humans.

2.4.2 Dynamometry

Isometric and isokinetic dynamometry are the most commonly used methods of investigating in vivo muscle function in humans, providing measures of muscle group force and force development (Gravel et al. 1990; Taylor et al. 1991; Aagard et al. 2002; Harrison et al. 2004; Young et al. 2005; Newton et al. 2006; Herda et al. 2008; Miyaguchi and Demura 2008; Bogaerts et al. 2009; O'Sullivan et al. 2009; de Ruiter et al. 2010; Jones and Bampouras 2010; Sekendiz et al. 2010; Edouard et al. 2011; Lanshammar and Ribom 2011; Stradijot et al. 2012; Engelen-van Melick et al. 2013; Menzel et al. 2013; Sanfilippo et al. 2013; Taylor and Fletcher 2013; Brown et al. 2014; Lew and Qu 2014; Yeung et al. 2014). Isometric dynamometry has been used by numerous authors in conjunction with ultrasound to assess the mechanical properties of tendon, in particular the patellar and Achilles tendons (Maganaris and Paul 2000; Maganaris et al. 2002; Maganaris 2003a; Reeves
et al. 2003; Maganaris et al. 2004b; Arampatzis et al. 2005; Narici et al. 2005; Reeves et al. 2005; Staﬁlidis et al. 2005; Albracht and Arampatzis 2006; de Monte and Arampatzis 2008; Morse et al. 2008; Ullrich et al. 2009; O’Brien et al. 2010; Peltonen et al. 2010). As a method, it is highly controlled but there are a number of issues which must be considered in relation to the accuracy, validity and reliability of measures obtained. Numerous authors have commented on the lack of validity of measures using the dynamometer, due to the absence of isometric and truly isokinetic contractions during dynamic sporting tasks (Abernethy et al. 1995). MTU loading is hence signiﬁcantly different to that experienced during typical sporting conditions, and the performance enhancing ability of the SSC is eliminated. However, in the absence of other reliable, controlled methods of assessing muscle group force capabilities, it is the preferred choice for assessment by healthcare professionals.

Measurement reliability is primarily dependent on the researcher using the identical protocol for every test session, but also on the level of familiarisation of the participants with the task especially for isokinetic testing. Most authors use at least one familiarisation session (Impellizzeri et al. 2008a; Hicks et al. 2013). When both the protocol is reliable and participants are sufﬁciently familiar with the task, dynamometry has been shown to a reliable method of assessing muscle strength at the hip and knee (Fenter et al. 2003; Maffiuletti et al. 2007; Tsiros et al. 2011) with all reporting ICCs of between 0.88 and 0.99. Chester et al. (2003) found reliability of healthy ankle measures to be slightly greater than those of injured patients, with ICC of between 0.74 and 0.92 reported for measures of plantarﬂexor torque. Moraux et al. (2013) reported an ICC of 0.88 with no evaluator effect for between-day isometric plantarﬂexion in 74 healthy participants. A signiﬁcant mean between-day difference of 5.46 N.m was reported, the equivalent of a 5% increase in strength. The limit of agreement was 30.6 N.m, suggesting that future reliability studies should also report these values as a LoA of 30.6 N.m is approximately a 30% difference.

Values for joint torque obtained from a dynamometer are composed of both joint torque and dynamometer compliance. Dynamometers assess strength as the magnitude of resistive torque which much be applied by the machine to resist the torque applied by the participant and maintain the joint angular velocity – the joint torque observed is a net torque, generated by both agonist and antagonist structures around a joint. The dynamometer is used to assess the strength of a muscle group, so ﬁxation of other body parts that may contribute to force output such as the trunk is necessary. To measure joint
torque requires alignment of the joint axis of rotation with the dynamometer axis of alignment (Croisier et al. 2005), and the magnitude of muscle contraction during this procedure is important. The timing of this alignment is reported by few researchers, but should be performed during a submaximal isometric condition when the joint is close to its position of maximum force capability (Baltzopoulos et al. 2012). Misalignment of the axes can induce large errors of between 2 and 17% into calculation of isometric joint torques (Herzog 1988; Kaufman et al. 1995; Arampatzis et al. 2004) and error of between 10 and 13% in isokinetic knee extensions (Kaufman et al. 1995). Errors of this magnitude may be greater than potential gains from a training protocol, which highlights how incorrect alignment may have serious consequences for observed changes. During isometric contractions, the joint angle should not change and should be recorded using goniometry. Due to the limited range of movement used by dynamometers and time required for it to reach the test speed, there is limited time for the muscle to develop maximum force at a prescribed speed (Baltzoloulos and Brodie 1989). The timing of when torque values are obtained is typically not reported in literature, but it is important that the researcher analyses the joint torque when the arm was moving at the velocity of interest (Baltzopoulos et al. 2012).

2.4.3 Force sledges

Force sledges have established themselves as a controlled yet dynamic method of assessing MTU function during exercise to gain insight into MTU behaviour during SSC tasks (Komi 2000). They have been used to investigate a variety of research questions, such as lower limb behaviour with post-activation potentiation (Comyns et al. 2006; Comyns et al. 2007; Flanagan and Comyns 2008), SSC function of healthy and anterior cruciate ligament patients (Flanagan and Harrison 2007; Flanagan et al. 2008), response to fatiguing exercise (Avela and Komi 1998; Strojnik and Komi 2000; Kuitunen et al. 2002a; Nicol et al. 2003; Harrison and Gaffney 2004; Kuitunen et al. 2004), effects of ageing on the lower limb (Harrison and Larkin 2006) and differences in SSC function due to sprint or endurance training (Harrison et al. 2004). More recent versions of this equipment have been designed to be light and as frictionless as possible, to try to make the jumps as reactive and natural as possible (Kramer et al. 2010). The control and fixed location of this method lends support to its use with invasive force measurement techniques such as optic fibres which feedback to simultaneous data logging equipment (Finni et al. 2000; Finni et al. 2001; Ishikawa and
Komi 2003; Ishikawa et al. 2006b; Sousa et al. 2007). The force sledge is an excellent mode of performing fatiguing SSC exercise due to the restriction of movement to one plane and the amount of control the participant has over their own movements (Ishikawa et al. 2006a; Nicol et al. 2006; Flanagan et al. 2009; Comyns et al. 2011).

Force sledge apparatus have been shown to be highly reliable in their measurement of force, force development and stretch-shortening cycle function, particularly during rebound jumps. The single and average ICC for peak vertical force were >0.907 and >0.967 for drop jumps and >0.953 and >0.984 during rebound jumps respectively. Vertical stiffness was >0.863 and >0.950 during drop jumps, and >0.907 and >0.967 during rebound jumps (Flanagan and Harrison 2007). Construct validity of this apparatus is well established, with differences in function observed between both old and young participants, and sprint and endurance trained athletes. Harrison and Larkin (2006) reported a very large effect size for age in measures of vertical stiffness, as they found differences of almost 3 kN.m\(^{-1}\) between young and old males of similar mass. Harrison et al. (2004) reported a difference of just over 2 kN.m\(^{-1}\) in vertical stiffness during a drop jump task and just under 2 kN.m\(^{-1}\) during a cyclical rebound task between sprint and endurance athletes. Kyröläinen and Komi (1995) reported similar training-related differences with concurrent increased muscle activation. The authors suggested that this increase in pre-activation aided in resisting the high impact loads in the eccentric phase of the SSC and aided in enhancing running economy.

Despite the strength of this apparatus as a reliable, controlled, valid method of investigating MTU behaviour during a stretch-shortening cycle task, it has never been used or adapted to study one muscle group in isolation. To the author’s knowledge, there is also no published literature which has combined it with motion analysis to study lower limb kinematics during an SSC task.

2.4.4 Oscillation

The oscillation technique is a method of assessing musculoarticular stiffness where the system is modelled as a linear damped mass-spring system and stiffness assessed based on the frequency response of a perturbed system (Ditroilo et al. 2011). In contrast to other methods, it is a global measure of the stiffness of a joint and its surrounding musculature, tendons, ligaments and other connective tissues. The measure is dependent on the
magnitude of loading rate, reflexive muscular characteristics, co-contraction, anthropometrics and fibre pennation angle. These factors are usually accounted for by standardising loading, joint angle, EMG and normalising to anthropometric measures such as segment length.

Due to the composite nature of the measure, its criterion validity against in vivo force and imaging techniques has not been established. The ecological validity of the task is also limited. Murphy (2003) reported very high between-day reliability for this method, with ICCs of between 0.89 and 0.95 and low coefficient of variation of between 5.4 and 8.5%. McLachlan et al. (2006a) suggests some familiarisation is required, as significant (9.57%) differences were reported between the first and second day of testing (ICC: 0.43, very large effect size of difference). Between-day results from day 2-3 were reliable, with similar values reported between day two and three (mean error: 3.15%, small effect size). Construct validity has been observed with increased stiffness associated with increased rate of isometric force development (Murphy 2003), increased concentric torque development during cycling, maximum isometric torque and isometric rate of torque development in the first 300 ms (Watsford et al. 2010). Males have also been shown to score higher in stiffness than females using this technique (Blackburn et al. 2006) further supporting the validity of the measure. Interestingly, decreased absolute and normalised (to mass) stiffness was observed in young females compared to postmenopausal females (Faria et al. 2011) despite an increased damping ratio in the younger females, suggesting there may be aspects of this measure more closely related to the tendon or muscle stiffness than the overall global measure.

2.4.5 Quick-release technique

The quick release method was originally reported by Goubel and Pertuzon (1973) to be related to the stiffness of the SE, with later work referring to this as a musculotendinous stiffness (Ochala et al. 2004; Ochala et al. 2005). Farcy et al. (2014) recently showed using ultra-high speed ultrasound that tendon was the primary contributor (72.5 ±9.5%) to the shortening of the MTU during the quick-release, followed by muscle fascicle (17.7 ±5.3%) and aponeurosis (9.7 ±8.6%). These percentage contributions were not dependent on the level of contraction or length of time of analysis. This method is particularly popular in continental European laboratories (Lambertz et al. 2001; Hof et al. 2002b; Lambertz et al. 2003; Rabita et al. 2008; Lapole and Perot 2011; Driss et al. 2012; Paiva et al. 2012;
Lambertz et al. (2013). During this test, participants are required to perform an isometric plantarflexion contraction at fixed percentages (e.g., 20/40/60/80%) of a previously established maximal voluntary contraction against a plate which is suddenly released. Torque (related to the isometric torque of the preceding 20 ms), angular displacement and velocity (then differentiated to obtain acceleration) are obtained during this protocol. Ochala et al. (2004) calculated stiffness as the ratio between the change in angular acceleration and the change in angular displacement, multiplied by the inertia value (dynamic torque divided by maximum acceleration). The slope of the stiffness-torque relationship is considered the stiffness index of the SE (Hof 1998).

Similar to the oscillation technique, the criterion validity of this method has not been established due to the composite nature of the measurement. Reliability was shown by Farcy et al. (2014) who reported ICC for musculotendinous stiffness between 0.83 and 0.96 and coefficient of variation of between 4.0 and 6.5% for the different level of contraction (30-80% of maximum voluntary contraction). These values are similar to those reported by Fouré et al. (2012).

Ageing, immobilisation, female gender and a non-athletic training background are all associated with lower muscle mass and magnitude of muscle available for activation, and decreased tendon stiffness. Supporting this construct, Rabita et al. (2008) reported stiffness of 1.83 N.m.rad⁻¹ for non-athletic controls and 2.76 N.m.rad⁻¹ for elite jumps athletes. Lambertz et al. (2001) reported no change in astronauts pre and post-space journey (4.83 ±0.16 rad⁻¹ vs. 4.78 ± 0.14 rad⁻¹) and Ochala et al. (2004) found a 55% increase in stiffness in older adults compared to young adults (3.14 ±0.45 rad.s⁻¹ vs. 2.03 rad.s⁻¹), which they attributed to altered muscle fibre composition. The inconsistent range of values observed and differences to other methods would suggest that further work in determining the construct validity of this method is required.

### 2.4.6 Electromechanical delay

Electromechanical delay (EMD) refers to the time difference between the onset of muscle activation and development of force in the muscle (Kubo et al. 2005b; Waugh et al. 2013). It is a composite measure of the time course of propagation of the action potential across the muscle membranes, calcium release from the sarcoplasmic reticulum and binding troponin, rate of cross-bridge formation and stretching of the series elastic element of the
muscle-tendon unit, with the last suggested to account for the majority of the measure (Cavanagh and Komi 1979). It can be obtained during both voluntary tasks and involuntary twitch techniques, which results in a shorter EMD (Zhou et al. 1995). Typical values for gastrocnemius and soleus using twitch techniques vary from <5 ms to 27.5 ±0.9 ms (Muraoka et al. 2004a; Nordez et al. 2009; Costa et al. 2010; Yavuz et al. 2010; Rampichini et al. 2013). As the focus of this thesis is on assessment of function during a task under voluntary control, this review will focus on values reported in this condition.

By its definition, EMD can only be assessed for one muscle at a time. Typical values for voluntary EMD of gastrocnemius are 22.8 ±8.2 ms during one legged-stance (Hopkins et al. 2007) and 26.6 ±6.0 ms and 32.6 ±7.6 ms for healthy limbs and those affected by Achilles tendinosis during a plantarflexion task (similar values were reported for lateral gastrocnemius). Values of between 26.9 ±2.4 ms and 27.4 ±2.9 ms were reported for healthy control males across a 4-8 week period which reduced from 27.8 ±2.2 ms to 24.4 ±2.2 ms with plyometric training (Wu et al. 2009). Winter and Brookes (1991) reported a time of 9.5 ±1.1 ms and 10.9 ±1.2 ms for males and females respectively, for what they referred to as a ‘force development time’ of soleus, which is what others refer to as EMD. In contrast, Wang et al. (2012) found soleus EMD to be 19.3 ±4.5 ms and 25.0 ±6.5 ms in healthy limb and those with Achilles tendinosis. This difference is most likely due to the test position, as Winter and Brookes tested participants with the knee and ankle at 90° whereas Wang et al. tested with the limb at 180°. With the knee at 90°, soleus is the sole ankle plantarflexor whereas with the knee extended, gastrocnemius has a larger contribution to force production.

When compared with other methods of assessing MTU function, the main strength of this method is the relative simplicity with which it can be used as there is little mathematical calculation involved and is a relatively simple set-up. The reliability of EMD measures is highly dependent on the reliability of the protocol used to measure EMG, and can be negatively influenced by the size of the muscle of interest. Almosnino et al. (2009) reported ICC of 0.23 to 0.79 for EMD of a number of neck muscles, with the most likely source of differences coming from the difficulty in accurately identifying an individual muscle as well as crosstalk (Chowdhury et al. 2013). The large surface area of the triceps surae allows for more accurate measurement of specific muscles, with Hopkins et al. (2007) reporting ICC of the triceps surae to be 0.972 for voluntary EMD during a plantarflexion movement. Glyn et al. (2009) also observed high levels of intra-subject intra-test reliability, with
coefficient of variation reported as between 3.1 and 6.5% for EMD of the dominant and non-dominant elbow flexors. Winter and Brookes (1991) reported test-retest coefficient of variation of 11.5% in males and 8.6% in females for measures of soleus EMD, and 15.7% in males and 13.3% in females for ECT.

The criterion validity of this method is not established. Construct validity as a measure of underlying musculotendinous stiffness has been shown by observed differences due to gender (Winter and Brookes 1991), training status (Clegg and Harrison 2005) and initial muscle length at which EMD is measured (Muro and Nagata 1985; Muraoka et al. 2004a), all of which are factors known to affect tendon stiffness using direct measures (Kubo et al. 2000a; 2000b; Maganaris et al. 2004b; Lichtwark et al. 2007b; Narici et al. 2008; Burgess et al. 2009a). Related changes in patellar tendon stiffness and vastus lateralis EMD have been reported by Kubo et al. (2001b) who observed an increase in EMD from 60 ±5.9 ms to 70 ±4.4 ms after 50 isometric contractions with a concurrent 22.7% increase in tendon compliance from 2.0 ±0.6 x 10^2 mm.N^-1 to 2.6 ±0.7 x 10^2 mm.N^-1. Kubo et al. (2001a) reported increased tendon stiffness from 67.5 ±21.3 N.mm^-1 to 106.2 ±33.4 N.mm^-1 and decreased EMD from 52.6 ±5.1 ms to 37.3 ±4.9 ms. Grosset et al. (2009) observed similar paired changes in musculotendinous stiffness assessed using the quick-release technique following endurance and plyometric training.

The lack of an assessment of criterion validity of the method still means that even though paired changes are observed, confounding variables such as muscle mass, or tendon architecture, biochemical changes or the amount of crimp in the tendon can be neglected. Harrison and Larkin (2006) reported decreased EMD in older adults, despite a large body of evidence using ultrasound that this measure decreases with age. Similarly, Kubo et al. (2005b) reported large differences in gastrocnemius EMD following cooling and heating (from 55.5 ±8.9 ms to 68.2 ±16.1 ms and from 54.4 ±6.1 ms to 63.1 ±11.4 ms) despite no observed change in Achilles tendon stiffness (consistently between 38.9 and 40 N.mm^-1). This suggests EMD is related to some measure of musculotendinous function but further study is required to identify what exactly this is.

2.4.7 Use of ultrasound

Ultrasound is a non-invasive imaging technique usually used in conjunction with some of the aforementioned methods such as dynamometry or the quick-release technique to gain
further insight into in vivo muscle and tendon behaviour during a variety of tasks. The most common areas imaged with ultrasound are the muscle-tendon junction (MTJ), muscle belly and transverse tendon to measure fascicle pennation angle, fascicle length, tendon elongation and subsequently tendon strain and stiffness. Elongation observed in the probe field of view is regarded as an expression of deformation of tissues distal to the measurement site (Bojsen-Moller et al. 2003). The simplest method to determine elongation is to calculate the distance moved by the muscle-tendon junction or a selected fascicle from an echogenic marker placed on the skin but this method is prone to error from excessive skin movement artefact and joint rotation.

The method of Lichtwark and Wilson (2005b) combining motion analysis with ultrasound is currently the only method which quantifies MTU behaviour in a truly ecologically valid test condition, as it is used during walking, hopping and running tasks (Lichtwark and Wilson 2005b; Lichtwark et al. 2007b; Farris et al. 2012). While the original work by Lichtwark and Wilson focused on the Achilles tendon, the method could be adapted for use on other tendons in the body. The muscle-tendon junction is digitised manually in the ultrasound probe field of view. Use of motion analysis system markers on the ultrasound probe allows for reconstruction of the probe local co-ordinate system and muscle-tendon junction location in the global co-ordinate system. Additional markers placed on the proximal calcaneus are an approximation of the Achilles tendon insertion point which allows for calculation of tendon length and tendon length change (Lichtwark and Wilson 2005b). Ideally, a second probe could be positioned over the calcaneal insertion to maximise measurement accuracy similar to that of Maganaris and Paul (2002), but due to the dynamic nature of the tasks that is not feasible.

Regression equations based on joint angles have previously been used to approximate muscle and tendon length (Ettema 1997; Herbert et al. 2002; Ishikawa and Komi 2004; Lichtwark et al. 2007a; Sousa et al. 2007; Lichtwark and Wilson 2008), with the equations of Grieve et al. (1978) and Hawkins and Hull (1990) the most popular. The length of aponeurosis can be calculated by assuming the muscle, tendon and aponeurosis and using ultrasound to quantify tendon length (Lichtwark and Wilson 2008). This also allows for calculation of strain and stiffness measures.

There are some associated issues when using ultrasound to assess MTU function. Use of manual digitising of the muscle-tendon junction may introduce a source of error into
tendon length measures (Bartlett et al. 2006). Within-digitiser reliability tends to be high as they operate to their own criteria for defining the muscle-tendon junction, fascicles or tendon outer border, but between-rater reliability can be poor for this reason (Ekizos et al. 2013). Digitising itself is also a slow, laborious and time-consuming process (Cronin and Lichtwark 2013), therefore the use of automatic tracking algorithms similar to those created by Loram et al. (2006), Lee et al. (2008), Cronin et al. (2011) and Gillett et al. (2013) is advantageous in processing this type of data.

2.4.8 Section summary

This section has provided an overview of the methods of assessing measures of plantarflexor function of interest, namely the force, rate of force development and mechanical measures. Each method has its own strengths and weaknesses related to validity (construct, criterion and ecological), control, reliability and simplicity, but there is a clear gap in the literature for a method that can reliably quantify these measures of the plantarflexors in a controlled yet dynamic, stretch-shortening cycle test condition.

2.5 Chapter Summary

Review of the literature shows the plantarflexors are an important muscle group to investigate due to their reported importance in everyday tasks such as hopping and running, high risk of injury, reported links to ACL injury and their role in modulating leg stiffness. Despite their importance, few existing measurement methods assess this muscle group during a SSC, which is the most relevant for everyday tasks. Hopping-based protocols tend to lack control which affects measurement reliability. This suggests a need for a controlled, reliable method of assessing plantarflexor function during a dynamic SSC task, which may help explain observed differences in limb function during a variety of tasks. Ideally, this method would allow for analysis of a number of measures of MTU function simultaneously, while being relatively simple to calculate. The following chapters will present a method for assessment of plantarflexor muscle function during a SSC task and application of this method to answer three related research questions.
Chapter 3: Reliability and consistency of plantarflexor stretch-shortening cycle function using an adapted force sledge apparatus
3.1 Abstract

The limitations of existing methods of studying plantarflexor stretch-shortening cycle function and muscle-tendon unit mechanics are predominantly related to measurement validity and reliability. This chapter describes an innovative adaptation to a force sledge which isolates the plantarflexors and ankle for analysis and the selection of a suitable analysis protocol. The aims of this study were to determine the sledge loading protocol to be used, most appropriate method of data analysis and measurement reliability in a group of healthy young adults. Twenty participants (11 males, 9 females; age: 23.5 ±2.3 years; height: 1.73 ±0.08 m; mass: 74.2 ±11.3 kg) completed 11 impacts at five different loadings rated on a scale of perceived exertion from 1 to 5, where 5 was a loading considered to be very hard to complete the 11 impacts using the adapted sledge i.e. an 11 repetition maximum. Analysis of impacts 4 to 8 or 5 to 7 using loading 2 (~70% of 11 repetition maximum) provided the most consistent results that were highly reliable (single ICC >0.85, average ICC >0.95), using similar kinematics to those observed in hopping and running. Results support use of an adapted force sledge apparatus as a valid, reliable method of investigating plantarflexor SSC function and MTU mechanics in a dynamic controlled environment.

3.2 Introduction

The SSC refers to a concentric contraction immediately preceded by an eccentric contraction, with the pre-stretch shown to enhance maximum work output of the muscle in the concentric phase (van Ingen Schenau et al. 1997b). It occurs repeatedly in movements like hopping or running (Bosco et al. 1981) which has made it the focus of much research (Finni et al. 2000; Geronilla et al. 2003; Ishikawa and Komi 2004; Regueme et al. 2005; Comyns et al. 2011). SSC function is related to both the force and power generated by the muscle and muscle-tendon unit (MTU) mechanics (Kubo et al. 1999). Of particular interest is the measurement of stiffness which refers to the ratio between force and linear or angular displacement. Stiffness can be measured at tissue, joint or limb level and its measurement has become increasingly popular due to its association with both performance and injury. It is believed that an optimal level of stiffness is required to store elastic energy and to absorb impact during landings (Bradshaw et al. 2006). Increased leg and joint stiffness is normally associated with improved performance in power activities (Kuitunen et al. 2002b; Harrison et al. 2004) possibly due to increased muscle cross-
sectional area (Ryan et al. 2009) and subsequent increased muscle activation. Excessive stiffness may increase the risk of bony injuries. In contrast, low levels of leg and tendon stiffness are typically associated with increased risk of soft tissue injury (Williams et al. 2004; Child et al. 2009; Arya and Kulig 2010; Maquirriain 2012).

The plantarflexors are important in power activities such as sprinting (Johnson and Buckley 2001; Bezodis et al. 2008) due to their role in force absorption and generation. These muscles and the ankle joint are particularly prone to injury during participation in sport (McManus et al. 2006; Fong et al. 2007; van Middelkoop et al. 2008; Fields et al. 2010). To date, there are few methods that can be used to reliably assess SSC function and MTU mechanics in a controlled yet ecologically valid environment.

Studies of in vivo plantarflexor SSC function in humans involve dorsiflexion-plantarflexion movements on a force plate (Kubo et al. 2000c) or dynamometer (Svantesson et al. 1991; Kubo et al. 2005a). The dynamometer improves control but lacks ecological validity for real-life activities such as hopping and running due to differences in kinematics and rate of force development, and unrealistic loading due to lack of impact events. Oscillation and quick-release techniques are often used to study the mechanical properties of the MTU. The oscillation technique involves modelling the limb as a linear damped-spring system. Oscillations in the force data are measured and input into formulae to determine musculoarticular stiffness. Low levels of perturbation suggest high levels of stiffness, and vice versa (Ditroilo et al. 2011). The quick-release technique measures the stiffness of the series elastic element of the MTU (Goubel and Pertuzon 1973). This method requires participants to perform an isometric plantarflexion contraction at a certain percentage of maximal voluntary contraction against a plate which is then suddenly released. Stiffness is calculated as change in joint moment divided by angular displacement (similar to joint stiffness) and related to the isometric torque of the preceding 20 ms with the stiffness-torque relationship considered an index of MTU stiffness. Reliability is established for both methods (Murphy 2003; McLachlan et al. 2006b; Lambertz et al. 2008) with construct validity established for the oscillation technique (Walshe et al. 1996). Neither method provides a measure of the stiffness of the tendon itself however, as measurement of that requires measurement of tendon elongation and tendon force.

Criterion valid methods of investigating MTU mechanics combine ultrasound imaging with isometric dynamometry (Reeves et al. 2003; Arampatzis et al. 2007; O'Brien et al. 2010) or
inverse dynamics (Lichtwark and Wilson 2005b). Construct and criterion validity of the
dynamometry method is well-established but content validity of this measure of tendon
stiffness is only representative of tendon loaded during isometric contraction. As most
measures of tendon mechanics are load-rate dependent (Paxton and Baar 2007), loading
during the preceding isometric contraction does not necessarily induce similar tendon
behaviour to SSC activities with cyclical eccentric and concentric contractions. The method
of Lichtwark and Wilson (2005) requires participants to hop on a force plate with an
ultrasound probe attached to the leg. This method has validity because it measures tendon
elongation and force but the reliability of ankle kinematics during a repetitive hopping
movement whilst the upper body is free to move can be poor (Joseph et al. 2013). This is
probably due to contributions from other body segments and uncontrolled movement. The
limitations of methods currently used to study plantarflexor SSC function and MTU
mechanics, in particular tendon stiffness, suggest a need for a valid (construct, criterion,
content and ecological) and reliable method of studying this muscle group.

Force sledges have previously been used to examine SSC function of the entire lower limb
(Ishikawa and Komi 2004; Comyns et al. 2011) and allow reliable, controlled measurement
in a dynamic environment. A typical lower body force sledge requires the participant to be
strapped into a seat as they move up and down on rails positioned at 90° to a fixed force
plate at an incline to the ground. This eliminates upper body contribution to lower limb
performance, while maintaining a similar movement pattern to hopping or running with a
series of impacts. Controlled impact velocity is one reason for improved reliability (Kramer
et al. 2010). This is thought to control eccentric muscle loading which can subsequently
affect concentric phase performance. Reliability of a lower-body sledge has been previously
established for peak force, leg stiffness, flight time and reactive strength index (Flanagan
and Harrison 2007). These variables provide important information about MTU loading
and function of the entire lower limb. There are several advantages to using a force sledge
for SSC and MTU mechanics studies that involve impacts, but due to the fixed force plate
and movable sledge design, the movement of the participant is necessarily large. Therefore,
force sledge systems present major difficulties when electromyography data or attachment
of an ultrasound probe may be required to obtain additional information on muscle or
tendon function.

The limitations of existing methods of studying plantarflexor SSC function and MTU
mechanics are predominantly related to measurement validity and reliability. This study
utilises an innovative adaptation to a force sledge which isolates the plantarflexors and ankle for analysis. With this adaptation, the force sledge is essentially reversed. The participant is fixed in place on a plinth at the base of the sledge and a foot contact plate is free to move up and down the sledge rails. The method is advantageous to other sledge protocols as the segment of interest remains in the one place yet the rhythmic loading of hopping and running is still replicated. This allows for the collection of ecologically valid kinematic, ultrasound and electromyography data with minimal noise artefacts. Plantarflexor loading can be increased with the addition of mass to the plate. However, there is a need to examine the reliability and internal consistency of the data produced by the adapted sledge design as a precursor to its use in examining SSC function and MTU mechanics. Therefore, the aim of this study was to determine the sledge loading protocol to be used, most appropriate method of data analysis and measurement reliability in a group of healthy young adults.

3.3 Methods

3.3.1 Participants

Following university ethics committee approval, a convenience sample of twenty participants (11 males, 9 females, age: 23.5 ± 2.3 years, height: 1.73 ± 0.08 m, mass: 74.2 ±11.3 kg) gave written informed consent to participate in this study. All participants were recreationally active, defined as participation in physical activity for at least 30 minutes a day 5 days per week. None had a history of lower limb surgery. They were also injury free in the lower limb for the preceding 3 months.

3.3.2 Participant preparation

12 mm retro-reflective markers were placed on the sledge plate edge, lateral 5th metatarsophalangeal joint (5MTP), lateral malleolus (MALL) and knee joint centre (KJC) of the preferred legs of participants. All trials were captured using a six camera 3D motion analysis system (500 Hz, MAC Eagle, Motion Analysis Corporation Inc., Santa Rosa, CA., USA). Participants were positioned at the base of the sledge as shown in Figure 7, and the thigh was secured using Velcro straps. Knee angle was maintained between 150º and 170º and hip angle was approximately 135º. Familiarisation consisted of approximately 25-30 impacts where the participant initially pushed the plate away from them and struck it rhythmically, followed by a second trial where the plate was released from 30 cm and the
participant was instructed to strike the plate as rhythmically as possible while minimising plate contact time. There was no mass added to the plate. This protocol was repeated until the participant was satisfied that they were familiar with the task and the researcher deemed they were striking the plate as instructed.

![Figure 7. Force sledge set-up, with participant in place and motion analysis markers attached.](image)

### 3.3.3 Test protocol

Testing commenced 5-10 minutes following familiarisation. All trials were completed using the preferred hopping leg. The plate began at a position 30 cm above the foot and was released after a ‘3, 2, 1’ countdown. Participants were instructed (as in familiarisation) to strike the plate in a marked area as rhythmically and continuously as possible for 11 impacts using solely the ankle joint, with the aim of minimising contact time while maximising flight time. Following each trial, the plate was secured away from the foot and participants were asked to rate their perceived effort from 1-5. These numbers corresponded to a difficulty rating of very easy (1), easy (2), OK (3), hard (4) and very hard (5) to complete the 11 impacts. Test trials were obtained at each of these perceived exertions. The maximum loading a participant could successfully complete 11 impacts with was achieved with as few trial efforts as possible to minimise fatigue effects. The loading order was as randomised as possible. A truly randomised order could not be used, as to use
a very heavy load early in the trial could potentially injure participants. The first and second trials hence used lighter loads.

Use of a one repetition maximum test to determine loading was unsuitable as the response of the plantarflexors during cyclical loading (i.e. multiple loadings) was of interest. Lichtwark and Wilson (2005) used the middle impacts of approximately 20 impacts in their analysis of tendon mechanics during a one-legged hopping task, so tests of multiple repetition maximum were considered more appropriate. The criteria used for selection of trials by these authors were not described. Use of increased number of trials is usually associated with increased reliability and decreased coefficient of variation (Hopkins et al. 2001), as well as providing insight into the natural variability of the participant (Mullineaux et al. 2001). Eleven impacts were chosen to provide a wide range of impacts for analysis and combinations of impacts for analysis within a single trial, without risking inducing unnecessary fatigue on participants. Pilot analysis showed the last impact was likely to be affected by motivation or participants losing count of the number of impacts, and Monaghan et al. (2007) found 10 trials to be more appropriate for reliable analysis of lower limb kinematics than protocols using two, four, six or eight trials. Use of an 11 impact test was hence considered appropriate.

Loadings were determined on the basis of perceived effort rather than absolute mass as no norms or guidelines currently exist for this apparatus. The rating of perceived effort has previously been shown to be related to the percentage of one repetition maximum in a variety of resistance exercises (Lagally et al. 2002). While use of an RPE may reduce the objectivity of the measure and be subject to training differences (Skatrud-Mickelson et al. 2011), it was considered the most appropriate and simplest approach to use as this is the first time this adaptation of the sledge has been used. A scale from one to five was used as it was the simplest for the participants to interpret and was a scale they were familiar with. During the 11RM test, the aim was to reach the maximum loading in as few efforts as possible. The use of a large range of values on a scale with numerous points was considered inappropriate due to potential excess trial efforts, which may subsequently induce participant fatigue. Even though the OMNI and Borg scales have been validated, they were not considered appropriate due to the range of values used and number of points (1-10, 6-20). The exercise was localised to one particular muscle group with little effect on other parts of the body. As a result, it was unlikely that heart rate would ever reach the level
of validated intensity during the test condition or that pictorial aid would help in determining perceived exertion.

Anchored scales have been proposed to provide a more accurate estimate of exertion, as participants are familiar with the range of feelings that may be experienced during a trial (Lamb et al. 2004). The frame of reference can be determined using either memory, verbal definition or experience (Noble and Robertson 1996). The scale used was semi-anchored as the low anchor (1) could be no lighter than the mass of the sledge on its own, and some of the lighter loadings were performed after their maximum effort (5; the high anchor) so they could compare those efforts to the highest intensity. Lagally and Costigan (2004) showed the type of anchoring procedure does not influence measurement reliability in resistance exercise.

Use of a loading equivalent to a fixed percentage of total body mass was inappropriate due to the localised nature of muscle loading and between-participant differences in plantarflexor muscle mass and strength. Similarly, use of a fixed loading equivalent to a percentage of maximal voluntary contraction obtained using a dynamometer was also considered inappropriate as contractions were not isometric or isokinetic. Mean sledge mass at each loading was 12.9 ±2.53 kg (loading 1: 55.9 ±5.9% of 11RM), 15.6 ±2.53 kg (loading 2: 68.2 ±5.3% of 11RM), 18.7 ±3.01 kg (loading 3: 80.2 ±5.2% of 11RM), 20.5 ±3.42 kg (loading 4: 89.3 ±3.7% 11RM) and 23.1 ±3.55 kg (loading 5: 100% 11RM). Participants were allowed at least 90 s rest between trials to allow recovery and reduce likelihood of fatigue (Read and Cisar 2001; Matuszak et al. 2003; Willardson 2006; Willardson and Burkett 2006) but if more time was required this was allowed. If participants were observed to be excessively moving their knee and contracting their quadriceps they repeated the trial and were reminded to use their ankle to generate force and not the knee. If this continued, testing ceased and the participant was removed from analysis.

3.3.4 Data treatment

Residual analysis of marker data was performed to find the cut-off frequency which provided the optimal balance between the amount of signal and noise present (Winter 2005). Residuals were calculated and plotted against cut-off frequency in Microsoft Excel (Microsoft Inc., Redmond, WA, USA), with the optimal cut-off frequency then
determined by hand. This was done on a marker-specific basis as the amount of noise in the data changed with marker location. Markers were filtered using a fourth order, zero lag, low-pass Butterworth filter with cut-offs of 12 Hz for the sledge and 5MTP markers, and 14 Hz and 18 Hz for MALL and KJC markers respectively.

The ankle angle was defined as the angle formed by the 5MTP, MALL and KJC in the sagittal plane, with 0° defined as an instance where the 5MTP, MALL and KJC were in a line. The angle tended towards 180° when the foot was in dorsiflexion. Ankle angle values were exported directly based on filtered data from motion analysis software (Cortex, Motion Analysis Corporation Inc., Santa Rosa, CA., USA). All other variables were calculated in Microsoft Excel (Microsoft Inc., Redmond, WA., USA).

The plate acceleration was calculated as the second derivative of plate position with respect to time, with force calculated using Newton’s second law. A correction for the component of weight acting down the sledge rails was used since the sledge was inclined at 30°. The formula used to calculate force was hence:

\[ F = m_s a_s + m_s g \sin 30 \]

Where \( m_s \) = total mass of the sledge plate, \( a_s \) = plate acceleration and \( g \) = acceleration due to gravity. Plate acceleration in free-fall due to gravity was predicted at 4.903325 m.s\(^{-2}\) (i.e. \( g \sin 30^\circ \)). The effect of friction was determined by calculating the mean measured plate acceleration in free-fall across 1500 data points from 10 participants in all 5 loadings. The difference between the calculated mean acceleration and the predicted acceleration at 30° inclination (4.903325 m.s\(^{-2}\)) was 0.00928 m.s\(^{-2}\) (0.189%), therefore friction was assumed negligible in subsequent calculations.

Peak force (\( F_p \)) was defined as the maximum force developed during each contact time with rate of peak force development (RPFD) calculated as the peak force divided by the time in seconds it took to reach it. Duration of eccentric loading was defined as the time period it took to reach maximum ankle angle, expressed as a percentage of contact time.

Examination of acceleration, velocity and displacement data for the adapted sledge system showed that a precise definition of the first contact “event” was not always clear from the acceleration data alone. In practice, the identification of non-zero acceleration appeared to be the best marker of foot contact on the plate. Therefore, contact time (CT) was defined as the period when plate marker acceleration was greater than zero and flight time (FT) defined as the period when it was zero or less. This is verifiable in comparison to the
velocity data which is non-linear during foot contact (Figure 8). Plate height (i.e. displacement from release to peak of flight) was calculated using the equations of motion and assumed the periods of upwards and downwards flight were equal. Reactive strength index (RSI) was defined as the ratio between plate height and preceding CT.

3.3.5 Data analysis

Five analysis strategies were implemented: 1) all impacts included in analysis, 2) impacts 2 to 11, 3) impacts 2 to 10, 4) impacts 4 to 8 and 5) impacts 5 to 7. Pilot work showed that the reliability and consistency of impacts 3 to 5 and 4 to 6 (different combinations of 3 consecutive impacts) were similar to those of impacts 5 to 7, therefore only the results of 5 to 7 are presented here. Each individual impact was included in each group for analysis. To account for temporal variation in foot contact times, the ankle angle data were normalised to contact time, with 0% representing the initial contact and 100% representing the release.

![Figure 8. Calculation of CT and FT based on sledge marker kinematics. CT was defined as the time period when acceleration was greater than 0 m.s\(^{-2}\) and FT as the time between consecutive CT.](image)

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3.3.6 Statistical analysis

Mean values and within-participant point-by-point standard deviation for all loadings and the five analysis strategies were calculated in Microsoft Excel. The mean within-participant, point by point standard deviation for all participants across the full CT was then calculated for each strategy and loading. Reliability, internal consistency and standardised variation of the variables was determined using two-way mixed measures single and average intra-class correlations (ICC) with absolute agreement (i.e. reliability), Cronbach’s α (i.e. internal consistency) and coefficient of variation (CoV; i.e. standardised variation) for each analysis strategy. ICCs and Cronbach’s α were calculated using SPSS Statistics 19 (IBM, Armonk, NY, USA). The most reliable loading was determined as that which resulted in the combined lowest CoV and highest Cronbach’s α and ICC.

![Figure 9. Ankle angle between-impact consistency for one participant using their preferred leg at loading 2. Solid line indicates mean ankle angle over the appropriate number of impacts, grey shading indicates point-by-point standard deviation.](image)

3.4 Results

The results showed that the adapted force sledge is a reliable method of investigating SSC function of the plantarflexors and the ankle joint complex. Similar results were observed for almost all variables for both preferred and non-preferred legs. Figure 9 shows data
from one participant at loading 2 using the preferred leg and a similar movement pattern was observed at all loadings. Maximum ankle angle was consistent within participants. The mean range of ankle angle was consistent between impacts for all loadings particularly when the analysis focused on only analysing five or three impacts with a visible reduction in maximum point-by-point SD. When focusing on three impacts rather than all eleven impacts, the range of mean point-by-point SD across a trial reduced by between 0.9 and 1.9 degrees for the various loadings. The largest reduction was in loading 2 where the mean point by point SD reduced from 4.9 degrees when analysing all impacts to 3.0 degrees when the analysis focused on impacts 5 to 7. Loading 2 also resulted in the lowest SD between impacts (3.0 degrees) with largest SD at loading 3 (3.5 degrees). The lowest reliability was reported in the duration of eccentric loading in the non-preferred leg, with high CoV even when analysing reduced numbers of impacts.

The results were similar for the SSC variables. A reduction in the number of impacts analysed resulted in decreased CoV and increased Cronbach’s 𝜋 and ICC in both legs. Depending on variable of interest, the use of a reduced number of impacts for analysis resulted in a reduction in the CoV of between 3.6% and 8.8%. Using this concentrated analysis, the Cronbach’s 𝜋 and average ICC for all variables at all loadings was greater than 0.90, suggesting high levels of reliability. Single measure ICC was highest for almost all variables at loading 2 for the preferred leg and was correspondingly high in the non-preferred leg. Tables 1 and 2 show mean, CoV, Cronbach’s 𝜋 and ICC for the variables of interest at the five loadings. The figures reported are when analysis focused on the middle impacts for both preferred and non-preferred limbs. Of note, the mean values obtained were dependent on the loading for $F_p$, RPFD, CT, FT and RSI.
Table 1. Reliability and consistency of variables at the five loadings during the middle impacts (preferred leg)

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Table 2. Consistency and reliability of variables at the five loadings during the middle impacts (non-preferred leg)

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### 3.5 Discussion

To our knowledge, this is the first study to use an adapted force sledge apparatus to examine plantarflexor SSC function in a controlled dynamic environment. Flanagan and Harrison (2007) showed the force sledge apparatus which allows for quantification of the entire lower limb's SSC function to be highly reliable with ICC (average measures) of 0.984, 0.987 and 0.989 and ICC (single measures) of 0.953, 0.963 and 0.968 reported for $F_p$, $F_T$ and RSI respectively. The high levels of consistency and reliability found for the adapted sledge apparatus and protocol used in this investigation have important implications for
future work, especially in relation to the measurement of MTU mechanics. The sledge protocol of Flanagan and Harrison (2007) allows for study of the entire lower limb as a whole but since the participant is attached to a moving chair, there are large movements of both participants and equipment. This may result in marker, electromyographic electrode or ultrasound probe movement and data which contain large amounts of noise. The adapted sledge apparatus used in this study ensures that the limb remains relatively stationary and is therefore useful for studies of the plantarflexors that require additional measures of MTU function. The isolation of the limb allows markers, electrodes or probes to be attached to the limb with minimal chance of movement thereby ensuring low noise signals. The results show low variability and high reliability of forces, RPFD and duration of eccentric loading, all of which may affect the measurement of MTU mechanics. The low levels of variability allow accurate detection of small changes over a period of time. This enhances the experimental sensitivity which can be important in experimental designs such as training interventions where changes over time may be quite small. This study used three-dimensional motion analysis which is considered the gold standard for marker kinematics to study the plate and ankle during cyclical loading. A potentially acceptable alternative to this method could involve the use of an accelerometer attached to the plate to measure acceleration and electrogoniometers attached to the ankle to measure ankle angles.

It is desirable to focus analysis on the middle impacts due to the reduced point-by-point SD of the ankle kinematics, suggesting increased movement consistency. CoV of the SSC variables also decreased when the analysis focused on these particular impacts. Cronbach’s $\alpha$ remained correspondingly high as did both single and average ICC suggesting this is an efficient analysis protocol for use with large groups. In studies of tendon stiffness, the measurement of tendon elongation requires time consuming manual digitisation of tendon ultrasound images. It is hence desirable to try to reduce the amount of data analysed while maintaining reasonable sample size and high reliability and consistency. Large decreases in the CoV were observed when the first and last impacts were removed from analysis, which is not surprising as the first impact is not part of the cyclical loading pattern. The last impact may be subject to cumulative fatigue or lack of concentration. The analysis strategies focusing on the middle impacts were more reliable as the participant was working in a consistent rhythm. These results are similar to Flanagan and Harrison (2007) who found measurement reliability during a rebound protocol to be higher than during a single
drop jump due to the rhythmical nature of the impacts. Use of the middle impacts also allows for tendon conditioning to occur before images are extracted for analysis, which is essential for accurate reliable measurement of tendon elongation and stiffness (Maganaris 2003b; Magnusson et al. 2008). As participants must undergo protocol familiarisation prior to testing, the tendon should have experienced enough conditioning prior to testing to ensure reliable measurement.

All loadings were reliable but use of loading 2 (approximately 70% of the 11RM) typically resulted in the highest single trial ICC and CoV scores. This loading resulted in good (ICC >0.85) to excellent (ICC >0.95) reliability for single and average measures. Additionally, the participants generally described this loading as being of low difficulty to complete 11 impacts. The use of heavier perceived loadings, particularly for a muscle group which rarely works in isolation, could result in early fatigue which is known to affect joint kinematics and kinetics (Sanderson and Black 2003; Kellis and Liassou 2009). The high levels of consistency within loading supports the use of a scale of perceived exertion as an appropriate method of determining sledge loading. The changes in mean values obtained for \( F_p \), RPFD, CT, FT and RSI must be considered in designing appropriate test protocols and prior to analyses, suggesting similar loadings must be used for comparison.

This is the first study to produce statistics to support the use of an adapted force sledge to increase control of the duration of eccentric loading on a muscle group or limb. The results show acceptably low standardised variability scores (CoV), high internal consistency (Cronbach’s \( \alpha \)) and high reliability (ICC) across all variables. Across all loadings, the mean duration of eccentric loading was consistently between 38 and 40% of CT, suggesting that sledge load was not a factor that affects phase duration. This is most likely because of the restriction of sledge and limb movement to one plane. This measure was more variable in the non-preferred limb most likely due to poorer motor control in the limb for dynamic tasks. The duration of eccentric loading is of particular interest in SSC studies, as researchers may wish to examine how manipulation of the eccentric phase affects subsequent concentric performance. Controlled loading is also beneficial in training studies which consider the effectiveness of eccentric training on MTU mechanics.

The method used in this study provided cyclical loading of the lower limb which is similar in contact times to real-life activities such as running and hopping (Babić et al. 2011; Hayes and Caplan 2012). The ankle joint kinematics and SDs using the sledge were also similar to
those reported in other studies (Mero and Komi 1985; Jacobs and van Ingen Schenau 1992; Hunter et al. 2005). The aim of this study was to determine the sledge loading protocol, most appropriate method of data analysis and measurement reliability in a group of healthy, non-injured participants for the application of the adapted sledge system. For simulation of cyclical loading patterns similar to hopping or running, the use of a subjective loading of 2 on a scale of perceived exertion ranging from 1 to 5 (equivalent to approximately 70% of an 11RM) and analysis of the middle (4 to 8 or 5 to 7) impacts provided consistent results that were highly reliable. In conclusion, the results show this apparatus is suitable for investigating plantarflexor MTU mechanics and SSC function and in a dynamic controlled environment.

3.6 Direction of future research in the context of the thesis

The force sledge has been shown to be a reliable, consistent method of investigating plantarflexor MTU and SSC function in a dynamic, controlled environment. Recent work has shown high levels of asymmetry in ankle kinetics during sprinting, but until now there has not been a method to quantify MTU function asymmetry during a SSC task. The aim of chapter 4 is hence to investigate differences in plantarflexor MTU and SSC function due to limb preference.
Chapter 4: Assessment of plantarflexor muscle-tendon unit function in healthy young adults - differences due to limb preference
4.1 Abstract

Most healthy humans move symmetrically at gross limb level but large kinetic and kinematic asymmetries have been observed at joint level during locomotion. The aim of this study was to assess muscle function asymmetries in healthy, active adults using an adapted force sledge apparatus which isolates the plantarflexors during a stretch-shortening cycle task. Peak force, rate of force development and stretch-shortening cycle function of preferred and non-preferred limbs were assessed in 21 healthy, active individuals using the adapted sledge and three-dimensional motion analysis. Between-limb differences and relationships were determined using paired t-tests/Wilcoxon-Signed rank test, Cohen’s d, absolute symmetry index and Pearson’s r/Spearman’s rho. Significant differences with moderate effect size (ES) were observed in peak force (ES: 0.66), rate of peak force development (ES: 0.78), rate of force development in the first 50 ms (ES: 0.76), flight time (ES: 0.64) and stretch-shortening cycle function (0.68), with no difference in contact time or duration of eccentric loading. A small ES (0.56) was observed in rate of force development in the first 30 ms. The upper range of asymmetry observed (up to 44.6%) was larger than previously reported for healthy individuals, indicating compensations occur at proximal joints during locomotion to ensure symmetrical movement.

4.2 Introduction

The assumption of limb symmetry in movement is reasonable as most healthy individuals walk and run without a limp or visible evidence of asymmetry (Flanagan and Harrison 2007). However, there is a growing body of evidence to suggest the legs behave differently in several kinematic and kinetic measures during dynamic tasks. An understanding of these asymmetries and establishment of the level of asymmetry in healthy individuals is important to avoid potential injury (Knapik et al. 1991) and aid in the monitoring of rehabilitation programmes. Knowledge of the kinematic and kinetic characteristics of the weaker or lesser skilled limb is also important in guiding skill development and enhancing performance (Ball 2011). Kinematic and kinetic asymmetries in healthy individuals have been observed at the ankle, knee and hip during walking and running tasks by several authors (Korhonon et al. 2010; Exell et al. 2012; Bredeweg et al. 2013). How the body coordinates these asymmetric and variable structures to produce movement is known as the degrees of freedom problem of movement coordination and control (van Emmerik et al. 2004). The traditional view is that variability and asymmetry represent biological noise.
but Davids et al. (2003) suggested these measures may be better viewed as the body’s response to the constraints imposed during performance and instead reflect large adaptive capacities and flexibility (Holt et al. 1995).

Asymmetry in lower limb function may be due to a variety of factors including lateralisation (Gabbard and Hart 1996), sport and training background (Cavanagh et al. 1977; Cromie et al. 2007), injury (Zifchock et al. 2006; Schiltz et al. 2009), speed of movement (Cavagna 2006; Schiltz et al. 2009), age (Atkins et al. in press), muscle activation (Valderrano et al. 2007) or contextual interference (Vagenas and Hoshizaki 1992). Quantification of limb function asymmetry and its underlying mechanisms is difficult for several reasons. It is a highly individualised and variable measure as people respond differently to greater demands on the musculoskeletal system (Smak et al. 1999) and any asymmetry observed is highly dependent on the test used (Jones and Bampouras 2010). As different tests of limb and muscle function give different values for asymmetry, assessments of between-limb differences must be valid for the task of interest. Tests must demonstrate similar contraction modes and kinematics as the task of interest to avoid erroneous detection of asymmetry which is not present during performance of dynamic tasks (Menzel et al. 2013).

The plantarflexors are a vital muscle group in locomotion, acting to both produce force and stabilise the shank upon landing (Kuitunen et al. 2002b). They act on the ankle joint, which is one of the most commonly injured areas of the lower limb in sport (Fong et al. 2007; Kolt 2013). Running and sprinting consist of repeated stretch-shortening cycles (SSC) i.e. cyclical eccentric contraction prior to concentric contraction. Previous work using a force sledge found no significant between-limb differences in peak force generation and SSC function of the entire lower limb in a cyclical loading task (Flanagan and Harrison 2007), but this may be due to compensatory motor control strategies rather than symmetrical behaviour at individual joint or muscle level. An upper limit of 15% asymmetry in healthy limbs has been previously suggested (Knapik et al. 1991), but there is no published research on isolated plantarflexor function in a SSC task to inform clinical decision-making on this.

Muscle function asymmetry assessment requires high levels of test validity, utilising similar contraction modes and range of motion as the task of interest. Furlong and Harrison (2013) developed a reliable, valid method of measuring plantarflexor muscle-tendon unit
(MTU) behaviour during a dynamic, fast SSC activity (i.e. contact time of less than 0.25 s). The task uses similar ankle kinematics and contact times as observed during running studies, with similar rates of peak force development reported as those observed in the Achilles tendon during walking (Finni et al. 1998). The aim of this study was to identify between-limb differences and level of asymmetry in plantarflexor muscle function of healthy, active adults during a dynamic, fast SSC task using this adapted force sledge. This may help to explain observed asymmetries in kinematic and kinetic parameters during gait, as well as indicate if compensation occurs in the lower limb during a cyclical task to ensure symmetrical movement.

4.3 Methods

4.3.1 Participants

Following institutional ethical approval, a convenience sample of 21 active individuals gave written, informed consent to participate in this study (10 females, 11 males; age: 23.8 ±2.27 years; mass: 72.5 ±11.9 kg; height: 1.72 ±.083 m). None had a history of lower limb surgery and all were injury free in the lower limb for the preceding 3 months. Participants were advised to refrain from unaccustomed strenuous activity for the 24 hours preceding data collection.

4.3.2 Test protocol: sledge test

An adapted force sledge was used to obtain measures of plantarflexor force, force development and stretch-shortening cycle function. A six camera 3D motion analysis system (500 Hz, MAC Eagle, Motion Analysis Corporation Inc., Santa Rosa, CA., USA) was used to track 9.5 mm retro-reflective markers on the sledge plate edge, the lateral 5th metatarsophalangeal joint (5MTP), malleolus (MALL) and estimated knee joint centre (KJC). Participants were positioned supine at the base of the sledge with the thigh secured at its proximal and distal end. They were instructed to strike the sledge foot plate in a marked area as rhythmically as possible, pushing it as high as they could while minimising contact time using only their ankle joint. Throughout this, the hip angle was approximately 135° and knee angle between 140° and 160°. Familiarisation consisted of approximately 25-30 impacts with no added mass where the participant initially pushed the plate away from them and struck it rhythmically and a second trial where the plate was released from 30 cm away from the foot. The protocol continued until the participant was satisfied that they
were familiar with the task and the researcher deemed they were consistently striking the sledge plate as instructed. Previous work has shown that subsequent trials at a variety of loadings resulted in low coefficient of variation and high intra-class correlation coefficients during each trial for the variables of interest (Furlong and Harrison 2013). This shows participants were sufficiently familiar with the task to produce reliable data even after 25-30 impacts.

The two limbs were tested sequentially, with the preferred leg familiarised and tested first as participants were more comfortable completing testing in this order. Pilot work showed the order of limb testing did not introduce bias into the data. Due to the nature of the task, the preferred leg was defined as the preferred leg for single-leg hopping. For all participants, this was also the preferred leg for kicking a ball. The sledge plate was released from 30 cm above the foot following a ‘3, 2, 1’ countdown. The same instructions were given as in familiarisation. The plate was secured away from the foot after successful completion of each trial and additional mass added to the sledge. Participants took a minimum of two minutes break between trials to minimise fatigue effects. If participants felt they needed longer to recover, this was allowed. The test was administered similar to an 11 repetition maximum (11RM) strength test, with the researcher attempting to reach the maximum loading in as short a time frame as possible. At least 90 impacts were completed per leg prior to data collection. Test load was approximately 70% of this 11RM (actual test load:17.4 ±3.9 kg, 24.4 ±6.2% body mass, 71.2 ±5.28% 11RM) as this has been shown to be the most reliable for the variables of interest (Furlong and Harrison 2013). The same protocol was used for the non-preferred leg with the same mass used for the test trial.

4.3.3 Test protocol: limb anthropometrics

As the effect of muscle size on force output is established, calf circumference to the closest millimetre was obtained using a flexible tape measure (Rollfix, Hoechstmass, Germany). Circumference was considered a simple, easily obtained measure which was correlated to plantarflexor muscle size (Bamman et al. 2000). The same experienced researcher obtained all measurements. Participants stood with the feet hip width apart and malleoli level on a 50 cm box so the researcher could obtain measures at eye level to avoid parallax error. Measures were obtained in the plane perpendicular to the long axis of the shank. Shank length was defined as the distance between the estimated knee joint centre and the lateral malleolus. Limb circumferences were recorded at a point immediately inferior to the
patella, at positions 25%, 50% and 75% from the proximal end of the segment, at the widest point of the calf and immediately superior to the malleolus. Five measures were recorded for each variable from each leg with the mean used for later analysis.

4.3.4 Data treatment and variable definition

Residual analyses were conducted to identify the optimum marker cut-off frequency and ensure an optimal signal: noise ratio (Winter 2005). Markers were filtered using a fourth order, zero lag, low-pass Butterworth filter with cut-off frequencies of 12 Hz for the sledge and 5MTP markers, and 14 Hz and 18 Hz for the MALL and KJC markers respectively. A visual check was used to ensure these cut-off frequencies did not over-smooth the impact event. Ankle angle was defined as the sagittal plane angle formed by the 5MTP, MALL and KJC markers, with the duration of eccentric loading defined as the percentage of contact time where the joint was dorsiflexing. Angles were exported directly from the motion analysis software (Cortex v3.1.1.1290, Motion Analysis Corporation Inc., Santa Rosa, CA., USA) with all other variables calculated in Microsoft Excel (Microsoft Inc., Redmond, WA., USA).

Plate acceleration was calculated as the second derivative of plate position with respect to time, with force calculated using Newton’s second law. A correction for the component of weight acting down the sledge rails was used since the sledge was inclined at 30°. The difference between calculated mean acceleration and predicted acceleration at 30° inclination during free-fall (4.903325 m.s\(^{-2}\)) has previously been calculated as 0.00928 m.s\(^{-2}\) (0.189%) therefore frictional force was considered to be negligible in calculations. Peak force (\(F_P\)) was the maximum force developed during each contact time with rate of peak force development (RPFD) calculated as the \(F_P\) divided by the time in seconds to reach \(F_P\). Rate of force development in the first 30 and 50 ms (RFD\(_{0-30}\), RFD\(_{0-50}\)) was calculated as the force value at 30 ms and 50 ms post-contact, divided by 0.03 or 0.05 s respectively. Contact time (CT) was defined as the period when plate marker acceleration was greater than zero and flight time (FT) defined as the period between consecutive CT. Plate height (i.e. displacement from release to peak of flight) was calculated using the equations of motion and assumed equal periods of upwards and downwards flight. The reactive strength index (RSI) refers to the ability to utilise the SSC, with increased RSI considered indicative of an increased ability to transition quickly from an eccentric to concentric contraction (Young 1995). It was defined as the ratio between plate height and preceding CT.
The middle impacts (5 to 7) were selected for analysis. Absolute symmetry index (ASI) was calculated as:

\[
ASI = \frac{|(X_P - X_{NP})|}{\frac{1}{2}(X_P + X_{NP})} \times 100\%
\]

Karamanidis et al. (2003)

Where \( X \) is the variable of interest and \( P \) and \( NP \) refer to the preferred and non-preferred legs respectively. The strength of this method is that it quantifies absolute difference between limbs rather than allowing positive and negative differences cancel each other out. A score of 0% indicates perfect symmetry between limbs.

4.3.5 Statistical analysis

Statistical analysis was completed in SPSS Statistics 20 (IBM, Armonk, NY, USA). The assumptions of normality and homogeneity of variance were assessed using Shapiro-Wilk’s test and Levene’s test. When a variable did not satisfy these assumptions, Wilcoxon’s signed rank test was used to determine if differences existed between the preferred and non-preferred leg. For all other variables, paired samples t-tests were used. Pearson’s r and Spearman’s rho were calculated to determine the strength of the linear relationship between the measures obtained from the two limbs. Correlation coefficients were considered large when the correlation coefficient was .5 to .7 and very large when between .7 and .9 (Hopkins 2006). The level of significance for all tests was set at \( \alpha < 0.05 \). Effect size was calculated using Cohen’s \( d_z \) using the formula

\[
d_z = \frac{m_z}{\sigma_z}
\]

where \( m_z \) = the mean of the observed between-limb differences and \( \sigma_z \) = the standard deviation of the observed differences (Cohen 1977). The scale for classification of effect size was 0.2 to 0.6 = small, 0.6 to 1.2 = moderate, 1.2 and above = large (Hopkins 2006). Where between-limb anthropometrics were significantly different, \( F_p \), \( RPF \), \( RFD_{0.30} \) and \( RFD_{0.50} \) were calculated normalised to the relevant circumference and between-limb differences and relationship recalculated.
4.4 Results

Non-parametric statistical tests were used to determine differences in CT, circumference superior to the malleolus, RSI and all absolute and normalised force and force development variables. Parametric statistical tests were used for all other variables. Ankle angle data was not obtained from three participants due to problems tracking the malleolus marker with the position of the sledge but for the remaining 18 participants, no significant difference was observed between legs in the duration of eccentric loading on the muscle (preferred leg: 39.1 ±4.2% of CT, non-preferred leg: 39.9 ±3.3% of CT, ASI: 9.4 ±8.1%, p = 0.546, negligible effect size). No differences were observed in CT (preferred leg: 0.168 ±0.033 s, non-preferred leg: 0.173 ±0.044 s, ASI: 9.5 ±8.8%, ASI: p = 0.433, negligible effect size) or most of the anthropometric measures. Limb circumference at the widest point of the calf was significantly different (preferred leg: 0.363 ±0.018 m, non-preferred leg: 0.361 ±0.019 m, ASI: 1 ±0.7%, p = 0.022, moderate effect size). Statistically significant differences in $F_p$, RPFD, RFD$_{0.30}$, RFD$_{0.50}$, FT and RSI were observed with large ASI from 11 ± 10.9 to 25 ±19.6%. Effect sizes were small for RFD$_{0.30}$ and moderate for the remaining measures. Significant differences with similar effect sizes and ASI were still present after normalisations to maximum calf width therefore only absolute values are presented in Table 3. Variability observed was almost the same magnitude as the mean observed value. All correlation coefficients were statistically significant, with large to very large linear relationships observed between the preferred and non-preferred limbs.

4.5 Discussion

The results lend strong support to the presence of asymmetry in measures of plantarflexor MTU function during a dynamic, controlled, SSC task. Asymmetries were observed in all measures of force, force development and SSC function without asymmetry in contact time or duration of eccentric loading, and most measures of estimated muscle size. Differences observed were hence due to differences in physiological and mechanical behaviour rather than altered kinematic strategies or estimated muscle size. The magnitude of difference in circumference of the two limbs was very small so it is not surprising that normalised data resulted in the same conclusions. Asymmetry in plantarflexor function has previously been observed (Valderrano et al. 2007) but the levels of asymmetry observed here (from zero to almost 45%) are greater than the 15% previously reported as the upper limit of asymmetry for healthy limb function (Knapik et al. 1991). This has implications for our understanding.
Table 3. Plantarflexor function differences with limb preference

<table>
<thead>
<tr>
<th>Variable (unit)</th>
<th>Preferred leg</th>
<th>Non-preferred leg</th>
<th>p-value</th>
<th>Effect size</th>
<th>Absolute symmetry index (%)</th>
<th>Between-limb correlation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak force (N)</td>
<td>553 ±186</td>
<td>499 ±136</td>
<td>.016</td>
<td>0.66</td>
<td>14 ±8.6</td>
<td>.76</td>
</tr>
<tr>
<td>Rate of peak force development (N.s⁻¹)</td>
<td>8448 ±3615</td>
<td>6976 ±2577</td>
<td>.004</td>
<td>0.78</td>
<td>23 ±18.8</td>
<td>.69</td>
</tr>
<tr>
<td>Rate of force development (0-30 ms) (N.s⁻¹)</td>
<td>11412 ±4663</td>
<td>10152 ±3239</td>
<td>.039</td>
<td>0.56</td>
<td>18 ±13.0</td>
<td>.69</td>
</tr>
<tr>
<td>Rate of force development (0-50 ms) (N.s⁻¹)</td>
<td>9972 ±3824</td>
<td>8680 ±2847</td>
<td>.008</td>
<td>0.76</td>
<td>17 ±14.8</td>
<td>.83</td>
</tr>
<tr>
<td>Plate flight time (s)</td>
<td>0.545 ±0.112</td>
<td>0.497 ±0.081</td>
<td>.008</td>
<td>0.64</td>
<td>11 ±10.9</td>
<td>.81</td>
</tr>
<tr>
<td>Reactive strength index</td>
<td>1.15 ±0.54</td>
<td>0.93 ±0.36</td>
<td>.005</td>
<td>0.68</td>
<td>25 ±19.6</td>
<td>.59</td>
</tr>
</tbody>
</table>

of natural variation in function between limbs, and recommendations for rehabilitation. It appears there is a greater range of acceptable asymmetry in measures of MTU function than previously thought as none of these participants proceeded to develop any plantarflexor or Achilles tendon problems in the 12 months following data collection. These results support the conclusions of Davids et al. (2003) who stated that even though measures of function may be considered abnormal when compared to the ideal medical model, they may still be healthy but reflect the adaptations of the limb to the constraints imposed during performance of the task. This large range also highlights the individualised nature of asymmetry.
The large correlations observed are of interest as the correlation coefficients for all measures of force, force development and RSI were greater than .59, i.e. for every 1 standard deviation change in the preferred limb there will be at least a .59 standard deviation change in the non-preferred limb (Hopkins 2006). This suggests that despite large differences the MTU still behaved similarly, i.e. force production in both limbs may be high, even if there is a difference between them. In healthy adults, this would be expected. Lower correlation may indicate injury or previous injury as the relationship between the healthy and injured limb may vary considerably. While the healthy limb may perform well on a task, the corresponding score from an injured limb may vary significantly dependent on extent of damage and rehabilitation.

The strength of this study is that it is one of the first to test plantarflexor muscle function asymmetries in a valid, dynamic test condition which other authors have highlighted as a major limitation of methods of assessing limb asymmetry (Menzel et al. 2013). Despite the high level of control afforded by dynamometry, it lacks similar loading patterns to SSC tasks in magnitude of forces, kinematics and impact. Accounting for a typical lever length of 16-18 cm from the malleolus to the ball of the foot, preferred force values observed here are greater than the concentric plantarflexor torque of 41.8 N.m observed during isokinetic dynamometry at 180 deg.s⁻¹ (So et al. 1994). This is most likely due to the increased involvement of the SSC during the task and increased time for force development. The results of this study may help explain the asymmetries observed by other authors during fast SSC activities. Exell et al. (2012) reported ankle joint work asymmetries of up to 93%. Joint work, as the time integral of joint power, is dependent on the product of joint torque and joint angular velocity which is subsequently dependent on speed of muscular contraction, rate of force development and opposing co-contraction. Altered joint torque may be the result of altered muscle group force production capabilities, and angular velocity can be determined by speed of muscle contraction, rate of force development and opposing co-contraction. However, previous work has shown altered proprioceptive capacities of the lower limb depending on whether alternating or bipedal hopping strategies are used (Travers et al. 2013). This may affect muscle pre-activation strategies prior to impact and potentially, observed force and force development values. Therefore, further study is required to identify how well the observed asymmetries in muscle function correspond to kinematic and kinetic asymmetries during activities such as running.
The human body can utilise a number of different degrees of freedom to produce the same output, and similarly utilise the same degrees of freedom coordinated differently to produce different output (Heiderscheit 2000). The results of this study clearly show a difference in output as assessed by plate flight time. These differences are most likely related to the differences observed in the force and force development capabilities of the plantarflexor muscle group as no differences were observed in the plate contact time or duration of eccentric loading on the muscle which are indicative of the strategy used by the participant to project the plate. Flanagan and Harrison (2007) found no significant between-leg differences in measures of peak force generated or RSI of the entire lower limb during repetitive cyclical loading. Similarly, Buckeridge et al. (2012) reported no significant differences in lumbar-pelvic kinematics during rowing despite observed between-leg differences in hip and knee kinematics, suggesting similar movements at the proximal limb do not reflect distal kinematics and kinetics at individual joint level. The results of this study show the plantarflexors of preferred and non-preferred limbs behave differently in their force-production capabilities, how quickly they develop force and the outcome of force application. This suggests the hip, knee and ankle joints of the lower limb compensate for each other to maintain overall lower limb symmetry during cyclical activities such as running. Previous work has shown differences in individual joint contributions to overall limb performance dependent on task demands (Hobara et al. 2011), so it is probable that the mechanisms of compensation also vary with task.

The idea of lower-limb compensation is in agreement with Exell et al. (2012) who reported large asymmetries in hip, knee and ankle kinematics and kinetics during maximal velocity sprinting. This supports previous findings suggesting the legs use different strategies to get off the ground (Sadeghi et al. 2000). Ball (2011) reported higher angular velocities at the distal joints of the preferred limb during kicking and lower angular velocities at the proximal joints compared to the non-preferred limb, which is consistent with Bernstein’s (1967) theory of motor control. This shows the ability of the two limbs to adapt differently to the conditions imposed and supports the idea of compensation in each limb to ensure successful performance. The regular preferential selection of one limb for performance of a dynamic task results in alterations in limb muscle function. During hopping or kicking, the preferred leg is used for a quicker movement with greater motor control and force development requirements while the non-preferred leg utilises more isometric muscular
contractions. This provides further evidence for the flexibility of the body to adapt to the constraints imposed to maintain symmetrical movement.

4.6. Direction of future research in the context of the thesis

Previous work has highlighted the need to assess limb function asymmetry in a valid, reliable test condition that uses similar contraction modes and kinematics as the task of interest. This chapter has used the adapted force sledge to investigate differences in plantarflexor function due to limb preference. The range of asymmetry is much greater than the 15% upper limit recommended by athletic trainers and healthcare professionals as the upper limit in healthy individuals. The wide range of asymmetry observed highlights the highly individualised nature of this particular measure, and indicates compensation may occur at the proximal joints to ensure symmetrical movement at gross limb level. These results may also help to explain the high levels of kinematic and kinetic asymmetry observed during running and sprinting.

The plantarflexors are particularly important in plyometric activities such as hopping and bounding, exercises which are frequently prescribed to both male and female athletes to enhance lower limb neuromuscular function. Chapter 5 investigates differences in plantarflexor function and mechanics due to gender during a plyometric task.
Chapter 5: Assessment of plantarflexor muscle-tendon unit function in healthy young adults - differences due to gender
5.1 Abstract

Plyometrics are high-intensity exercises commonly used by the strength and conditioning coach to improve performance in stretch-shortening cycle (SSC) activities and prevent injury. This training is effective due to the rapid development of forces within the muscle and alterations in the mechanical properties of the muscle-tendon unit, both of which are known to influence performance in SSC activities. To date, it is unknown if a gender-related difference exists in the plantarflexor response to plyometric loading. An adapted force sledge was used with three-dimensional motion analysis to investigate the response of the plantarflexors of 34 age and training-matched males and females to cyclical impact. Contact times, flight times and reactive strength index were found to be similar between groups. Statistically significant differences in absolute peak force and rates of force development of up to 55% were observed during loading. With normalisation, small and moderate effect sizes were observed for all force and force development variables but only peak force remained statistically significant. Absolute stiffness, peak eccentric and concentric power and work were statistically significant with moderate effect size, but with normalisation only peak eccentric power and work were statistically significant with a moderate effect size. The results of this study have implications for our understanding of the mechanical properties of the plantarflexors of the male and female athlete and may help to explain, in part, the increased risk of knee injury in females during landing tasks.

5.2 Introduction

Plyometric training utilises stretch-shortening cycle (SSC) muscle actions to enhance the ability of the muscle to produce maximal force as rapidly as possible (Markovic and Mikulic 2010). Typical plyometric exercises are bounding and hopping based, such as drop jumps, hurdle jumps and single leg hops. Their effectiveness in enhancing performance in SSC activities such as running, sprinting and jumping, both alone and in combination with other training modalities, is well established (Delecluse et al. 1995; Markovic 2007; Impellizzeri et al. 2008b; Markovic and Mikulic 2010). As a result, strength and conditioning coaches frequently incorporate these exercises as part of training programs for athletes in a wide range of sports. Their importance in improving lower limb neuromuscular control has also led to widespread use by coaches and athletic trainers as part of anterior cruciate ligament (ACL) injury prevention programs (Mandelbaum et al. 2005; Hewett et al. 2006a; Myer et al. 2006; Myer et al. 2008).
The most effective training interventions have a high level of specificity to the performance task of interest both in terms of muscle group used and muscle loading. The plantarflexors have been shown to be extremely important in the acceleration and maximal velocity phases of sprinting (Johnson and Buckley 2001), acting as both major force producers and limb stabilisers (Kuitunen et al. 2002b). Maintenance of lower limb stiffness during plyometric training activities is vital to maximise training effectiveness. Previous work has shown the ankle joint to be an important contributor to lower limb stiffness during hopping tasks (Farley and Morgenroth 1999) and with increased demand for shorter ground contact time, gastrocnemius and soleus electrical activity increases (Hobara et al. 2007). Effective plyometric training is dependent on the utilisation of consecutive fast SSCs (fSSC) where the ground contact time is less than 0.25 s (Schmidtbleicher 1992), with sprinting one of the best examples of a fSSC activity (Cavanagh and Lafortune 1980; Paavolainen et al. 1999; Kuitunen et al. 2002b). The combination of these factors may explain, in part, why this training modality is so effective in enhancing sprint performance and neuromuscular control.

One of the aims of plyometric training is to enhance SSC function through maximal utilisation of the eccentric phase of contraction of the muscle-tendon unit (MTU). SSC function is affected by musculotendinous mechanics (Anderson 1996). Plyometric training has been shown to enhance neuromuscular function of the lower limb through increased neural drive, improved muscular coordination and altered mechanical properties and architecture of the MTU (Markovic and Mikulic 2010). These alterations can also improve force absorption capacity in impact situations. A gender difference in muscle activation strategies to maintain lower limb performance has previously been observed during two-legged jumps in place (Padua et al. 2005). The mechanical properties of the plantarflexor MTU have previously been reported to vary with gender (Blackburn et al. 2006; Hoge et al. 2010). Females have been shown to exhibit a larger decrease in Achilles tendon stiffness compared to males following stretching (Burgess et al. 2009a) as well as lower musculoskeletal stiffness during isometric dynamometry (Granata et al. 2002a). Decreased stiffness in females during a hopping task was observed by Granata et al. (2002b) but could be explained by lower body mass. While this suggests stiffness modulation is influenced by body mass, a decreased ability of females to effectively utilise the SSC in the upper body has also been observed (Miyaguchi and Demura 2009). Consideration of these results would suggest that the male and female MTU behaves differently during SSC tasks, with
only some differences attributable to differences in body mass. Previous work has shown that males and females utilise different strategies during single leg landings (Weinhandl et al. 2010) which may be due, in part, to underlying differences in MTU physiological and mechanical functional capacities (Mayhew and Salm 1990; Perez-Gomez et al. 2008). Athlete gender may hence be an important factor the strength and conditioning professional needs to consider in selection of exercises for optimum program design.

Recent work has developed an adapted force sledge based method of isolating the plantarflexors for analysis during a fSSC task (Furlong and Harrison 2013). It has been shown to be highly reliable and controlled, while maintaining validity as a dynamic fSSC task. Use of this adapted sledge could provide important information about gender differences in the plantarflexor response to impact. The aim of this study was to identify differences in plantarflexor force, force development, SSC function and mechanical behaviour in males and females during cyclical loading. This could have important practical applications in understanding the effective design of plyometric, fSSC training programs.

5.3 Methods

5.3.1 Participants

Following university ethical approval, 17 males (age: 23.2 ±2.73 years; height: 1.78 ±0.05 m; mass: 80.8 ±7.81 kg) and 17 females (age: 24.2 ±1.87 years; height: 1.68 ±0.07 m; mass: 65.4 ±6.90 kg) gave written informed consent to participate in this study. The male and female groups were matched for age within 18 months and training background. Each group of 17 participants contained seven field sports athletes on similar training programs, eight recreational athletes and two sprint-trained athletes. All volunteers participated in physical activity for at least 30 minutes a day 4 days per week. None had a history of lower limb surgery or were currently taking any medications. They were injury free in the lower limb for the preceding 3 months and were advised not to participate in any unaccustomed strenuous exercise for the 24 hours preceding data collection.

5.3.2 Test protocol

An adapted force sledge apparatus consisting of a wooden plate free to move along rails angled at 30° to the horizontal was used to determine plantarflexor function during the fSSC task. This apparatus has been shown to be highly reliable in its measurement of
plantarflexor force and force development, both for single (ICC >0.85) and average (ICC >0.95) measures (Furlong and Harrison 2013). A 9.5 mm marker placed on the side of this plate was tracked by three three-dimensional infra-red motion analysis cameras (500 Hz, MAC Eagles, Motion Analysis Corporation Inc., Santa Rosa, CA., USA). Participants were positioned supine at the base of the sledge with the thigh secured to a solid support at its proximal and distal ends. The ankle joint was free to move to push the plate up the inclined rails. Knee angle was between 140° and 160° during impact.

The preferred hopping leg was used for analysis. Participants were instructed to strike the plate rhythmically in a marked area pushing it as high and as fast as they could, using only the ankle joint. Participants were initially asked to push the plate away from them using the above criteria then the plate was dropped towards their foot from a 30 cm height after a ‘3, 2, 1’ countdown. This protocol was repeated for a minimum of 25-30 impacts until the participant was satisfied that they were familiar with the task and the researcher judged them to be striking the plate as instructed.

A similar protocol to that of a repetition maximum test was used, with the ultimate aim to achieve the maximum loading a participant could exert for 11 impacts. Participants were allowed two minutes rest in between trials but if more time was required this was allowed. Similar to familiarisation the plate began at a position 30 cm above the foot and was released after a ‘3, 2, 1’ countdown. Participants were instructed to continue striking the plate for a total of 11 impacts. Following each trial, the plate was secured away from the foot. A loading equivalent to 70% of this 11RM was used for the test trial as this has previously been shown to be highly reliable and a loading that participants were comfortable with using.

### 5.3.3 Data treatment and variable definition

Markers were filtered using a fourth order, zero lag, low-pass Butterworth filter. Residual analysis was conducted to identify the optimum cut-off frequency to ensure an optimal signal: noise ratio (Winter 2005), with a cut-off frequency of 12 Hz used for the sledge marker. A visual check was also performed to ensure the impact was not excessively smoothed by use of this cut-off frequency. Filtering was completed in Motion Analysis software (Cortex v3.1.1.1290, Motion Analysis Corporation Inc., Santa Rosa, CA., USA). All variables were calculated in Microsoft Excel (Microsoft Inc., Redmond, WA., USA).
The 5th to 7th impacts were considered representative of the plantarflexor response to cyclical SSC loading, and have previously been shown to be consistent and reliable for analysis of the variables of interest (Furlong and Harrison 2013). Plate acceleration was calculated as the second derivative of plate position with respect to time, with force calculated using Newton’s second law with a correction for the component of weight acting down the sledge rails as the sledge was angled at 30°. Contact time (CT) was defined as the period when plate marker acceleration was greater than zero and flight time (FT) defined as the period when it was zero or less. These definitions were used as when the sledge was away from the foot it was in free-fall therefore the only forces acting on it were gravity (−4.905 m.s\(^{-2}\)) and friction. Friction during free-fall was calculated as 0.18% so was considered negligible for these calculations. Peak force (\(F_p\)) was the maximum force developed during each contact with timing of peak force (%CT) referring to the percentage of CT where peak force occurred. Rate of peak force development (RPFD) was calculated as the peak force divided by the time in seconds it took to reach it. Rate of force development in the first 30 and 50 ms (\(RFD_{0-30}\), \(RFD_{0-50}\)) was calculated as the force at 30 and 50 ms, divided by 0.03 and 0.05 s respectively.

The stiffness of the plantarflexor MTU (\(K_{PF}\)) was calculated as \(F_p\) divided by the peak displacement of the sledge during impact in the direction of the sledge rails. Power was calculated as the product of sledge force and sledge velocity (measurement unit: watts, W), with work done calculated as the time integral of power (measurement unit: joules, J). The eccentric phase was defined as the time period when power was negative, and concentric phase defined as when power was positive. Peak power (\(P_{ECC}\), \(P_{CONC}\)) and total work done (\(W_{ECC}\), \(W_{CONC}\)) were calculated for each phase. Plate height (i.e. displacement from release to peak of flight) was calculated using the Galileo equations of motion, assuming equal periods of upward and downward flight. Reactive strength index (RSI) was defined as the ratio between plate height and preceding CT.

The influence of increased muscle size on force production is established therefore all force-related variables were normalised to total plantarflexor muscle volume by dividing observed force measures by personalised total plantarflexor muscle volume. This was calculated using the simple regression equations of Handsfield et al. (2014) for soleus, gastrocnemius medialis and lateralis, peroneus longus and brevis, tibialis posterior, flexor hallucis longus and flexor digitorum longus and based on height and mass products. Volume was considered a suitable variable to normalise to due to its relationship to muscle
physiological cross-sectional area (PCSA) which has an established relationship with force-producing capacity (Powell et al. 1984). It has been used previously in the literature to account for differences in size between individuals, particularly in tasks using isolated muscle groups (Morse et al. 2005b). PCSA could not be used due to the technical difficulty in obtaining personalised muscle fibre lengths and pennation angles for each participant. Both absolute and normalised values are presented in the results.

5.3.4 Statistical analysis

The mean absolute and normalised value for each variable from the three impacts for each participant was used for input into statistical analyses. Mean and standard deviation for each gender group were calculated for each variable to provide normative values and estimate of sample variability. Groups were considered well matched as similar variability was observed in each group. All statistical analysis was completed using SPSS Statistics 20 (IBM, Armonk, NY, USA). Between-group differences were determined using independent samples t-tests. The assumptions of this test were checked using Shapiro-Wilk’s test of normality and Levene’s test of equality of variances. Where data violated the assumptions, differences were determined using a Mann-Whitney U test. Level of significance was set at $\alpha < 0.05$. Effect size (practical significance) was calculated using Cohen’s $d$ using the formula

$$d = \frac{|m_M - m_F|}{\sigma}$$

where $m_M$ and $m_F$ are the means of the male and female groups respectively, and $\sigma$ is the pooled standard deviation of the two groups (Cohen 1977). The scale for classification of effect size was 0.2 to 0.6 = small, 0.6 to 1.2 = moderate, 1.2 and above = large (Hopkins 2006). Moderate and large effects were considered practically significant.

5.4 Results

Parametric statistical tests were used for FT and normalised values for $F_p$, RpFD, $RFD_{0.50}$, $W_{ECC}$ and $W_{CONC}$. Non-parametric tests were used for all other variables. During cyclical loading, no practical or statistically significant between-group differences were observed in the manner in which participants struck the plate (as defined by CT or %CT where $F_p$ occurred). CT was indicative of a fSSC, with CT of $0.183 \pm 0.038$ s and $0.174 \pm 0.035$ s observed in males and females. This difference was less than 5% and not significant. The
timing of \( F_p \), \( P_{ECC} \) and \( P_{CONC} \) were similar for males and females with non-significant differences of negligible and small effect size observed.

Statistically significant differences in calculated total plantarflexor muscle volume of large effect size were observed (1086 ±113 cm\(^3\) for females, 1360 ±122 cm\(^3\) for males). Statistically significant between-group differences with moderate (trending towards large) or large effect sizes were observed in all absolute measures of force and force development (\( F_p \), \( RPFD \), \( RFD_{0.30} \), \( RFD_{0.50} \)). With normalisation, differences were small to moderate in size but only the difference in \( F_p \) was still statistically significant (Table 4).

Statistically significant differences of moderate effect size were observed in absolute measures of stiffness, peak power and work in both eccentric and concentric phases (Table 5). With normalisation, differences in \( K_{pf} \) became negligible. Similarly, \( P_{CONC} \) and \( W_{CONC} \) were no longer statistically significant and of small effect size. \( P_{ECC} \) and \( W_{ECC} \) remained statistically significant and of moderate effect size.
Table 4. Differences in plantarflexor force, force development and flight time due to gender

<table>
<thead>
<tr>
<th>Variable (unit)</th>
<th>Absolute values</th>
<th>Normalised to plantarflexor muscle volume in cm³</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Females</td>
<td>Males</td>
</tr>
<tr>
<td>Plate flight time (s)</td>
<td>0.502 ±0.072</td>
<td>0.560 ±0.114</td>
</tr>
<tr>
<td>Peak force (N)</td>
<td>447±86</td>
<td>661 ±167</td>
</tr>
<tr>
<td>Rate of peak force development (N.s⁻¹)</td>
<td>6360 ±1603</td>
<td>9856 ±3350</td>
</tr>
<tr>
<td>Rate of force development (0-30 ms) (N.s⁻¹)</td>
<td>8925 ±2594</td>
<td>13271 ±4344</td>
</tr>
<tr>
<td>Rate of force development (0-50 ms) (N.s⁻¹)</td>
<td>7755 ±1699</td>
<td>11577 ±3510</td>
</tr>
<tr>
<td>Variable (unit)</td>
<td>Absolute values</td>
<td>Normalised to plantarflexor muscle volume in cm³</td>
</tr>
<tr>
<td>---------------------------------</td>
<td>-----------------</td>
<td>-----------------------------------------------</td>
</tr>
<tr>
<td></td>
<td>Females</td>
<td>Males</td>
</tr>
<tr>
<td>Stiffness (N.m⁻¹)</td>
<td>7957 ±2470</td>
<td>10403 ±3429</td>
</tr>
<tr>
<td>Peak eccentric power (W)</td>
<td>286 ±80</td>
<td>486 ±194</td>
</tr>
<tr>
<td>Eccentric work (J)</td>
<td>31 ±10</td>
<td>50 ±17</td>
</tr>
<tr>
<td>Peak concentric power (W)</td>
<td>246 ±73</td>
<td>393 ±191</td>
</tr>
<tr>
<td>Concentric work (J)</td>
<td>34 ±11</td>
<td>50 ±20</td>
</tr>
<tr>
<td>Reactive strength index</td>
<td>0.89 ±0.27</td>
<td>1.19 ±0.58</td>
</tr>
</tbody>
</table>
5.5 Discussion

No significant differences were observed in CT or %CT which shows both groups utilised the fSSC and the muscle group was loaded for a similar duration. The results of this study show that within an age and training background matched sample of healthy young adults, moderate to large differences exist in the plantarflexor force, force production and mechanical behaviour of males and females during a cyclical loading task. While some differences were removed with normalisation to muscle volume, a number of other important measures remained both statistically and practically significant.

The lower absolute $F_p$, RPFD, $\text{RFD}_{0.30}$ and $\text{RFD}_{0.50}$ observed in females is similar to previous work but this study appears to be one of the first to establish this in a dynamic, fSSC task. With normalisation, these differences became smaller but still of moderate effect size. There are a number of reasons why males achieved higher scores in force and force development tasks. Males have been shown to have greater muscle thickness and increased pennation angle than females in both the soleus and gastrocnemius muscles (Chow et al. 2000), both of which increase force production capacity and may account for differences observed. Difference in calculated muscle volume was statistically and practically significant with a large effect size. This suggests muscle size can partly explain observed differences but as not all differences were removed with normalisation, other factors must also affect observed measures. Males are known to be more powerful due to increased testosterone and/or an improved ability to effectively utilise the SSC (Miyaguchi and Demura 2009). The SSC is thought to be most effective with increased tendon stiffness and males have been shown to have a stiffer Achilles tendon than females of similar training background probably due to a lack of circulating oestrogen and increased muscle mass. Increased tendon stiffness shifts the force-velocity curve to the right so for the same muscle contraction velocity, force production capabilities are higher. This could explain, at least in part, why higher forces and rates of force development were observed in male participants.

Absolute rates of force development during both tasks were significantly lower in females. With normalisation, these differences were still of moderate effect. Differences in RPFD and $\text{RFD}_{0.50}$ were of greater effect than $\text{RFD}_{0.30}$, possibly due to increased contribution of muscle activation in these variables and the decreased ability of females to develop force rapidly. These results are of particular interest in relation to the role of the plantarflexors in prevention of serious knee injury. Several authors have suggested the plantarflexors may
have a role to play in the prevention of ACL injuries (Fleming et al. 2001; Elias et al. 2003; Mokhtarzadeh et al. 2013) and as mentioned previously plyometric training has been used with success in ACL injury prevention programs. Gastrocnemius has been shown to be an antagonist of the ACL increasing anterior tibial translation (Fleming et al. 2001) but soleus has been shown to exert posterior forces on the tibia due to its anatomical origin (Elias et al. 2003). As the physiological cross-sectional area of soleus is approximately 3-8 times larger than the gastrocnemius (Fukunaga et al. 1992), this suggests the plantarflexors as a group will act to reduce ACL loading. ACL injuries are most likely to occur 17 to 50 ms following initial ground contact (Krosshaug et al. 2007). If a female has a decreased ability to generate force which will act to protect the ACL in that period of time, then this gender related deficit in rate of force development may place females at an increased risk of injury. While this may be due in part to muscle activation differences, muscle size must also be considered. Resistance training with an emphasis on muscle hypertrophy may be desirable to reduce the risk of potential injury, but the decreased capacity for hypertrophy in females must also be considered.

A statistically significant difference in absolute $K_P$ was observed, which was not present with normalisation. This is similar to the findings of others (Granata et al. 2002a; Granata et al. 2002b; Padua et al. 2006), and highlights the importance of muscle size and volume in the modulation of stiffness. Maintenance of a stiff ankle joint is an important coaching point in performance of plyometric activities. The plantarflexors are important in the modulation of ankle and leg stiffness, especially during short contact hopping. Similar to the measures of force and force development, establishment of increased muscle size through resistance training may be desirable to maximise plyometric training effectiveness. No difference was observed in normalised $P_{CONC}$ or $W_{CONC}$, which is in contrast to previous work which found increased concentric power in males during a countermovement jump due to increased muscle activation (Caserotti et al. 2001). During the eccentric phase, muscle activation has less of an influence which explains the lack of observed difference in that phase. The results of this study show eccentric power and work to be lower in females even with normalisation, due in part to the smaller forces generated. These differences are hence also due to previously reported differences in tendon stiffness, fibre size and/or pennation angle in females. Decreased $P_{ECC}$ indicates a decreased ability to respond in situations requiring high forces and rapid execution such as landing (Caserotti et al. 2001) and decreased ability to dissipate force (LaStayo et al. 2003). Females are known to land
from plyometric exercises such as drop jumps using a more upright landing technique and increased levels of plantarflexion. While the knee is mainly responsible for force absorption, the next largest energy absorber in females are the plantarflexors (Decker et al. 2003). Females have been shown to demonstrate altered neuromuscular function during these tasks as well as possessing mechanical, biochemical and anatomical risk factors which predisposes them to injury in landing (Hewett et al. 2006b; Park et al. 2009; Pollard et al. 2010). These results add further evidence for the increased risk of injury for females during these tasks. These results are particularly relevant to the strength and conditioning professional. The decreased ability of the female plantarflexor MTU to absorb force suggests that to avoid risk of injury, female athletes should commence plyometric training at a lower intensity than male counterparts, e.g. drop jumps should be performed from a lower initial starting height.

Males tend to have larger body and muscle mass than females, so scaling of data to account for differences in body size is necessary to allow us determine if meaningful differences truly exist (Batterham et al. 1997). Normalisation to body mass or a power of body mass is commonly used for scaling of forces in tasks such as jumping or running as the entire body has an influence on the force output (Mullineaux et al. 2006). Caution has been advised in relation to the use of body mass as the normalisation variable in isolated muscle tasks as there is no definitive relationship between increased body mass and increased specific muscle mass (Nevill et al. 2004). As a result, force for isolated muscle groups is often normalised to physiological cross-sectional area (Morse et al. 2007) or muscle volume (Morse et al. 2005b). While normalisation to physiological cross-sectional area is acknowledged as the most appropriate method to scale muscle forces, it is very difficult to accurately measure muscle volume, muscle fibre length and muscle pennation angle without expensive, highly technical equipment and methods (Akagi et al. 2009). Muscle volume is most accurately determined by MRI or CT but this facility is not freely available for use. The regressions of Handsfield et al. (2014) based on height and mass products can account for between 21 and 66% of individual plantarflexor muscle volumes. While there is still a large proportion of unaccounted variance, their method appears to be a better variable to normalise to rather than solely body mass. Their findings support the inconsistent relationship between body mass and individual muscle volume as they found equations based solely on body mass performed worse in the prediction of muscle volume. There are some limitations to its use in this study, such as its generic, non-gender-specific
nature. Due to the typical differences in body composition for males and females, even within similar activity levels it is possible some errors may be present and that gender-specific regression analyses may be more suitable for prediction of muscle volumes. Whether allometric scaling or a ratio method should be used is also debatable (Jaric et al. 2002) and there were a number of highly active individuals within this study so their predicted muscle volumes may be prone to errors. Use of a linear normalisation of force to muscle volume was considered to be the most appropriate analysis for this set of data as the definitive answer to these problems is still an area of developing research.

No significant difference was observed in measures of RSI. This result is initially surprising as MTU mechanics which are known to be affected by gender (Blackburn et al. 2006; Hoge et al. 2010) influence SSC effectiveness. Previous work has also demonstrated gender differences in RSI during jumping tasks (Ebben et al. 2009). However, as no differences were observed in CT or FT this result is to be expected. The use of the same relative load for both groups may also explain the lack of observed difference. It is possible that if the same absolute load was used for both groups, a difference may have been observed due to the decreased strength ability of the females which would subsequently influence CT and/or FT. This result highlights one of the limitations of the use of RSI by the strength and conditioning professional, as similar RSI was observed despite differences in the force, rate of force production and mechanical behaviour of the plantarflexor MTU. These results suggest caution is advised in the use of RSI as a tool to monitor training intensity and effectiveness as has been suggested previously (McClymont 1995; Flanagan and Comyns 2008). More direct measure of MTU behaviour using electromyography and ultrasound imaging may provide us with more information on how the male and female MTU behaves differently during this particular task.

5.6 Direction of future research in the context of the thesis

This chapter has shown differences in plantarflexor function and mechanical properties due to gender in a group of age and training-matched healthy young adults. Investigation of ankle joint and Achilles tendon mechanical properties would provide further insight into the function of the plantarflexors during a SSC task, but the sledge rails cannot take the weight of a force plate. Chapter 6 investigates the use of a motion analysis marker-based method of determining the centre of pressure for potential use in inverse dynamics analyses.
Chapter 6: A marker-based method of determining centre of pressure during a hopping task
6.1 Abstract

The structure of the adapted force sledge does not allow for use of a force plate to determine the centre of pressure (CoP) during impact. To date, errors reported using alternative methods to the force plate for determining CoP during dynamic tasks have been high. The main aim of this chapter was to investigate the accuracy of a motion analysis marker-based system to determine CoP during a hopping task through a number of analyses. The potential assumption of a fixed CoP during impact was shown to be inappropriate but the positions of a number of markers attached to the feet were shown to be highly correlated to CoP position during two legged hopping. Five markers were attached to the left and right feet of eight volunteers (5 females, 3 males, age: 25.0 ±2.8 years, height: 1.75 ±0.07 m, mass: 71.3 ±11.3 kg). Multivariate forward stepwise and forced entry linear regression was used with data from five participants to determine CoP position during quiet standing and hopping task at various frequencies. Maximum standard error of the estimate of CoP position was 12 mm in the anteroposterior direction and 8 mm in the mediolateral. Cross-validation was performed using the remaining three participants. Maximum root mean square difference between the force plate and marker method was 14 mm for mediolateral CoP and 20 mm for anteroposterior CoP during 1.5 Hz hopping. Differences reduced to a maximum of 7 mm (mediolateral) and 14 mm (anteroposterior) for the other frequencies. The smallest difference in calculated sagittal plane ankle moment and timing of maximum moment was during 3.0 Hz hopping, and largest at 1.5 Hz. Results indicate the marker-based method of determining CoP may be a suitable alternative to a force plate to determine CoP position during a two-legged hopping task at frequencies greater than 1.5 Hz.
6.2 Introduction

Centre of pressure (CoP) refers to the point of application of the ground reaction force which is normally acquired using a force plate (FP). CoP position, position change and magnitude of area within which it moves are of interest to researchers during both standing and dynamic tasks such as hopping, walking and running (Han et al. 1999; Lafond et al. 2004; Hertel et al. 2006). It is also used as an input for inverse dynamics calculations of joint torques (McCaw and Devita 1995). The fixed location of most FPs has led researchers to attempt to find more mobile alternatives to determine CoP position such as in-shoe devices (Chesnin et al. 2000; Forner-Cordero et al. 2004; Fradet et al. 2009). However, due to large reported differences between the FP and these methods, the FP still remains the most commonly used method of obtaining CoP.

Root mean square difference (RMSD) between CoP determined using a FP and in-shoe measurement systems has previously been reported to be between 15 mm during quiet standing and 41 mm during walking (Chesnin et al. 2000; Fradet et al. 2009). Differences are most likely due to insole movement, incorrect transformation of co-ordinates from local to global systems or temporal synchronisation error (O'Connor et al. 1995). Pillet et al. (2010) used a motion analysis anthropometric-based model to determine CoP in static and self-selected speed walking tasks. RMSD was between 14.2 ±5.2 and 17.6 ±5.7 mm in the mediolateral (ML) direction and 33.0 ±4.2 mm and 43.4 ±5.7 mm in the anteroposterior (AP) direction. Shifts in CoP of these magnitudes have been shown to significantly affect sagittal plane joint torque calculation by 19.5% to 48.2 % (McCaw and Devita 1995; Kim et al. 2007).

The main aim of this chapter was to investigate potential methods of determining the CoP during a dynamic task. This was achieved through four objectives. Firstly, the accuracy of an assumption a fixed centre of pressure during hopping was investigated. The second objective was to investigate if markers attached to the feet had any relationship with the CoP position. The third was to determine the accuracy of a motion analysis marker-based system to determine CoP during a hopping task, by comparing values obtained with those obtained from a FP. The final objective of this chapter was to investigate the effects of using this surrogate method on ankle joint torque calculations. The results of this chapter have applications in both clinical and practical sports settings, enhancing the mobility of accurate measurement due to reduced need for the fixed location of a FP.
6.3 Methods

6.3.1 Assumption of a fixed centre of pressure during hopping

6.3.1.1 Participants

Four healthy active participants (age: 26.7 ±2.36 years, height: 1.73 ±0.06 m, mass: 69.6 ±7.6 kg) volunteered for this pilot study. All were healthy and injury free for at least 3 months prior to testing. All were familiar with performance of two-legged hopping but a brief familiarisation prior to testing consisted of at least ten hops at each test frequency (1.5 Hz, 2.2 Hz, 3.0 Hz) until the participant was striking the FP in time with the metronome (TempoPerfect Metronome Software v2.02, NCH Software, Canberra, Australia).

6.3.1.2 Test protocol

Two-legged hops were performed with one foot on each force plate (1000Hz, AMTI OR6-7, AMTI Inc., Watertown, MA., USA) and hands on hips. FP X and Y axis was aligned to the ML and AP directions respectively. A minimum of 50 impacts per participant per trial was collected. Participants were allowed as much rest time as required between trials to minimise fatigue. The FP was re-zeroed between each trial to minimise drift.

6.3.1.3 Data treatment and analysis

All data was exported to Microsoft Excel (Microsoft Inc., USA) with each impact isolated for analysis. Data was not filtered as pilot work had shown the original force trace to be quite clean. Filtering the forces made minimal difference (<0.1%) to the magnitude of the force data. Each impact was normalised to 101 data points using a custom-written LabVIEW programme (National Instruments Inc., TX, USA). Range of movement, standard deviation and 95% confidence interval (95% CI) of CoP movement were calculated for each frequency in Microsoft Excel.

6.3.2 The relationship between marker kinematics and CoP

6.3.2.1 Participant

One healthy active female participant (age: 24 years, mass: 64.1 kg; height: 1.71 m) participated in this pilot study.
6.3.2.2 Test protocol

9.5 mm retro-reflective markers were placed on the superior aspects of the first and third metatarsophalangeal joints (1MTP and 3MTP), the lateral aspect of the 5th metatarsophalangeal joint (5MTP), at half the length of the foot at a point in line with 3MTP (midfoot, MF) and on the superior foot at the point where it joined the leg (groove, GR). Data was acquired as described in section 6.3.1.2.

6.3.2.3 Data treatment and statistical analysis

Marker kinematics were filtered using a fourth order, reverse pass, low pass Butterworth filter at a cut-off frequency determined by residual analysis (Winter 2005). Marker positions in the X, Y and Z axes and CoP position were exported from the motion analysis software (Cortex v3.1.1.1290, Motion Analysis Corporation Inc., Santa Rosa, CA., USA) for analysis in SPSS Statistics 20 (IBM, Armonk, NY, USA). Pearson’s correlation was calculated between all variables to identify relationships between CoP position and calculated marker kinematics. Correlation coefficients were considered small when r was between .1 and .3, moderate when r was between .3 and .5, large when r was between .5 and .7 and very large when between .7 and .9 (Hopkins 2006).

6.3.3 Regression analysis

6.3.3.1 Participants

Following university ethics committee approval, a convenience sample of eight healthy, active participants (5 females, 3 males, age: 25.0 ±2.83 years, height: 1.75 ±0.07 m, mass: 71.3 ±11.3 kg) consented to participate in this study. Retro-reflective markers (9.5 mm) were attached to the same anatomical landmarks identified in 6.3.2.2: 1MTP, 3MTP, 5MTP, MF and GR. Familiarisation consisted of at least ten hops at each test frequency (1.5 Hz, 2.2 Hz, 3.0 Hz and a self-selected frequency) until participants were striking the FP in time with the digital metronome (TempoPerfect Metronome Software v2.02, NCH Software, Canberra, Australia).

6.3.3.2 Test protocol

All trials were captured using a six camera 3D motion analysis system (300 Hz, MAC Eagle, Motion Analysis Corporation Inc., Santa Rosa, CA., USA) and two AMTI force plates (300 Hz, AMTI OR6-7, Watertown, MA., USA). This sample rate was considered appropriate as
the purpose of data collection was to obtain simultaneous marker and CoP position rather than investigate changes in CoP position or force with time. Testing consisted of two two-legged hopping trials of 30 seconds duration at each test frequency with one foot on each FP and during quiet standing. The two-legged hopping movement was similar to that performed by Farley and Morgenroth (1999) and Hobara et al. (2010), where the two feet were positioned hip width apart and both legs jumped simultaneously in place. FP X and Y axes were aligned to the ML and AP directions respectively. The FP was re-zeroed between every trial to minimise drift but some trials required re-zeroing in post-processing to produce a zero force value when nothing was on the plate.

6.3.3.3 Data treatment

Markers were filtered using a fourth order, zero lag, low-pass Butterworth filter in the motion analysis software with cut-offs of 14 Hz (1MTP, 3MTP), 13 Hz (5MTP) and 17 Hz (MF, GR) based on residual analyses (Winter 2005). CoP data calculated using vertical forces of less than 100 N and occurring in the outer 10 cm edges of the FP were removed from analysis due to previously reported inaccuracies in CoP measurement at low force levels and the outer edges of the FP (Bobbert and Schamhardt 1990; Middleton et al. 1999). Only ML CoP data occurring within the width of the foot were included in analysis due to the improbability of the CoP being outside the foot when the foot was in contact with the plate at high forces.

6.3.3.4 Error checking

A calibrated mass (9.815 kg) was placed on the FP close to the centre of the plate, then to the right, left, behind and front of this position for 30 s each to estimate CoP deviation when an inanimate object was placed on it. The influence of zeroing in post-processing was estimated by comparing the same trial twice with the trial zeroed at different points.

6.3.3.5 Statistical analysis

All statistical analysis was completed using SPSS Statistics 20 (IBM, Armonk, NY, USA). CoP data from five participants were used for initial equation derivation. Multivariate forward stepwise linear regression ($p_{in} = 0.05, p_{out} = 0.10$) was used to derive frequency-specific equations for predicting CoP position from marker position in the X, Y and Z axes. The most important markers to predict CoP in ML and AP directions were identified with consideration to prevalence in equations and practicalities of marker location.
Co-ordinate data from all three axes for these markers were then entered into multivariate forced entry linear regression (Table 9). All input data were marker position in metres. To clarify presentation, standard error of the estimate and cross-validation results are presented in millimetres.

6.3.3.6 Cross-validation

Adjusted $r^2$ was calculated during forced entry regression to provide an estimate of explained variance in the population. Cross-validation was completed using data from the remaining three participants to determine how well the model predicted CoP in similar participants from outside the sample. RMSD between CoP measured using the FP and the marker-based method was calculated for these participants. 95% CI of the difference between the two methods for all eight participants was calculated to provide an estimate of the predictive ability of the equations. Pearson’s $r$ was calculated to investigate the strength of the linear relationship between the two methods with the correlation considered very large when between .7 and .9, and ‘nearly perfect’ when greater than .9 (Hopkins 2006).

6.3.4 Effect on joint torque calculation

6.3.4.1 Participants

The data of six participants (age: 24.2 ±2.37 years, mass: 73.4 ±12.99 kg; height: 1.76 ±.093 m) from the same data collection as that described in 6.3.3 were used for analysis. Three participants were selected at random from this group to be analysed during each frequency, with each participant analysed for at least two frequencies.

6.3.4.2 Test protocol and analysis

The same protocol and set-up as 6.3.3 was used therefore will not be repeated again in this section. Additional 9.5 mm retro-reflective markers were placed on the lateral malleolus of the participants and filtered at 14 Hz. Three participants were chosen at random for analysis at each frequency. Joint moments were calculated using both FP CoP and marker-based CoP with the equations of Winter (2005) and the segment inertia model of Dempster (1955). All analysis was calculated in a custom-written Microsoft Excel template (Microsoft Inc., USA) which was validated against manual calculations completed by hand and a custom-written Mathcad programme (PTC Corp., Needham, MA, USA) used for calculating inverse dynamics. As horizontal force during the impact on the force sledge
tends to be zero due to the sledge design, a third analysis where horizontal force was assumed to be zero was also calculated. Joint moments calculated using the FP CoP and regression-based CoP were compared on an impact-by-impact basis using a custom-written Visual Basic (Microsoft Inc., USA) macro. RMSD of calculated moments across the entire impact and differences (mean, absolute and as a percentage of maximum) were determined between the maximum moment calculated using the FP CoP and the moment using the regression method at the corresponding instant, and between the maximum moments using both methods. Mean and standard deviation for each of these variables across all impacts for all participants was then calculated.

6.4 Results

6.4.1 Assumption of a fixed centre of pressure during hopping

A general trend of decreased movement and movement variability was observed with an increase in hopping frequency (Table 6). CoP movement was much smaller in the 3.0 Hz hopping condition.

6.4.2 The relationship between marker kinematics and CoP

Very large to almost perfect linear relationships were observed between marker position and CoP position during two-legged hopping, with \( r = .804 \) to \( .990 \) observed for all marker positions in the related axis (e.g. 1MTP\(_x \) and CoP\(_x \)). Moderate relationships (\( r < .448 \)) were observed between marker position in the two other respective axes and CoP position. During hopping at 2.2 Hz, \( r \) of \( .691 \) (1MTP\(_z \)) and \( .708 \) (3MTP\(_z \)) were found with measures of CoP\(_x \) under the left foot.
Table 6. Centre of pressure movement during two-legged hopping (all values in mm)

<table>
<thead>
<tr>
<th></th>
<th>1.5 Hz</th>
<th>2.2 Hz</th>
<th>3 Hz</th>
</tr>
</thead>
<tbody>
<tr>
<td>Range of movement (ML)</td>
<td>43</td>
<td>43</td>
<td>36</td>
</tr>
<tr>
<td>Standard deviation ML CoP</td>
<td>15</td>
<td>20</td>
<td>14</td>
</tr>
<tr>
<td>95% confidence interval ML CoP</td>
<td>29</td>
<td>39</td>
<td>27</td>
</tr>
<tr>
<td>Range of movement (AP)</td>
<td>65</td>
<td>58</td>
<td>37</td>
</tr>
<tr>
<td>Standard deviation AP CoP</td>
<td>14</td>
<td>12</td>
<td>14</td>
</tr>
<tr>
<td>95% confidence interval AP CoP</td>
<td>27</td>
<td>23</td>
<td>28</td>
</tr>
</tbody>
</table>

6.4.3 Regression analysis

The number of data points included in the analysis are shown in Table 7. During the static plate trial, CoP location was shown to deviate by between 0.4 and 0.7 mm in both X and Y axes, dependent on the position of the mass on the plate. Post-zeroing the plate affected CoP by 2 to 3 mm in the X axis and 1 mm in the Y.

Use of a two marker model explained high levels of variance (91.4% to 99.9%) and resulted in small SEE (1 to 14 mm, Table 8). A full model improved predictive ability slightly with 92.8% to 99.9% explained variance, and smaller SEE (1 to 11 mm). The prevalence of the 1MTP and MF markers and necessity of the 5MTP marker in inverse dynamics analysis formed the basis of selection for markers in the forced entry regression equations.
### Table 7. Number of data points used in multivariate forward stepwise and forced entry regression, and cross-validation

<table>
<thead>
<tr>
<th>Frequency (Hz)</th>
<th>Input for</th>
<th>Mediolateral (left)</th>
<th>Mediolateral (right)</th>
<th>Anteroposterior</th>
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<tbody>
<tr>
<td>Quiet standing</td>
<td>Regression</td>
<td>5355</td>
<td>13277</td>
<td>18632</td>
</tr>
<tr>
<td></td>
<td>Cross-validation</td>
<td>4691</td>
<td>2144</td>
<td>6835</td>
</tr>
<tr>
<td></td>
<td>Regression</td>
<td>4804</td>
<td>10452</td>
<td>15256</td>
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<td>1.5</td>
<td>Cross-validation</td>
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<td>3340</td>
<td>4994</td>
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<tr>
<td></td>
<td>Regression</td>
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<td>4682</td>
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<td>Regression</td>
<td>4698</td>
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<td>3235</td>
<td>3787</td>
<td>7022</td>
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<tr>
<td>Self-selected</td>
<td>Regression</td>
<td>6382</td>
<td>6586</td>
<td>12968</td>
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<td></td>
<td>Cross-validation</td>
<td>11778</td>
<td>3598</td>
<td>15376</td>
</tr>
</tbody>
</table>

Forced entry regression resulted in similar accounted variance as the full 5 marker set in the forward stepwise regression (Table 9). SEE was 1 to 8 mm (ML) and 3 to 12 mm (AP). 95% CI in the ML direction ranged from 4 mm in quiet standing up to 18 mm in 1.5 Hz hopping. In the AP, values ranged from 16 mm in 3 Hz hopping to 28 mm in 1.5 Hz hopping. Almost perfect relationships (r ≥ .95) were observed between the two methods for all equations, except for ML CoP of the right foot during 1.5 Hz hopping which was very strong (.87).
Table 8. Initial multivariate forward stepwise regression analysis to identify main predictors of centre of pressure position in mediolateral and anteroposterior directions

<table>
<thead>
<tr>
<th>Frequency and direction</th>
<th>Predictors used</th>
<th>$R^2$</th>
<th>SEE (mm)</th>
</tr>
</thead>
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<tr>
<td>Quiet standing (left)</td>
<td>MF$_X$, GR$_Y$</td>
<td>.999</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>5MTP$<em>X$, MF$</em>{XZ}$, GR$_X$</td>
<td>.999</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Full model</td>
<td>.999</td>
<td>1</td>
</tr>
<tr>
<td>Quiet standing (right)</td>
<td>MF$_{XZ}$, GR$_Z$</td>
<td>.997</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>3MTP$<em>Z$, MF$</em>{XZ}$, GR$_Z$</td>
<td>.997</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Full model</td>
<td>.998</td>
<td>1</td>
</tr>
<tr>
<td>Quiet standing (anteroposterior)</td>
<td>3MTP$_Z$, 5MTP$_Y$</td>
<td>.937</td>
<td>7</td>
</tr>
<tr>
<td></td>
<td>3MTP$<em>Z$, 5MTP$</em>{YZ}$, MF$_{YZ}$</td>
<td>.959</td>
<td>6</td>
</tr>
<tr>
<td></td>
<td>Full model</td>
<td>.989</td>
<td>3</td>
</tr>
<tr>
<td>1.5 Hz mediolateral (left)</td>
<td>1MTP$_X$ and GR$_X$</td>
<td>.987</td>
<td>5</td>
</tr>
<tr>
<td></td>
<td>1MTP$_{XZ}$, 5MTP$_Z$ and</td>
<td>.993</td>
<td>3</td>
</tr>
<tr>
<td></td>
<td>Full model</td>
<td>.994</td>
<td>3</td>
</tr>
<tr>
<td>1.5 Hz mediolateral (right)</td>
<td>1MTP$_X$ and 5MTP$_X$</td>
<td>.960</td>
<td>6</td>
</tr>
<tr>
<td></td>
<td>1MTP$_X$, 5MTP$_X$ and MF$_Z$</td>
<td>.965</td>
<td>6</td>
</tr>
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<td></td>
<td>Full model</td>
<td>.985</td>
<td>4</td>
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<tr>
<td>1.5 Hz (anteroposterior)</td>
<td>MF$_{YZ}$, 1MTP$_Y$</td>
<td>.959</td>
<td>14</td>
</tr>
<tr>
<td></td>
<td>MF$<em>{YZ}$, 1MTP$</em>{YZ}$, 3MTP$_X$</td>
<td>.964</td>
<td>13</td>
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<tr>
<td></td>
<td>Full model</td>
<td>.972</td>
<td>11</td>
</tr>
<tr>
<td>2.2 Hz mediolateral (left)</td>
<td>1MTP$_Z$ and 3MTP$_X$</td>
<td>.914</td>
<td>9</td>
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<tr>
<td></td>
<td>1MTP$<em>Z$, 3MTP$</em>{XZ}$, 5MTP$_Z$</td>
<td>.920</td>
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<td>Full model</td>
<td>.928</td>
<td>8</td>
</tr>
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<td>Description</td>
<td>Measurement</td>
<td>R</td>
<td>N</td>
</tr>
<tr>
<td>-------------------------------------------------</td>
<td>-----------------</td>
<td>----</td>
<td>---</td>
</tr>
<tr>
<td>2.2 Hz mediolateral (right)</td>
<td>1MTP&lt;sub&gt;x,y,z&lt;/sub&gt;, MF&lt;sub&gt;x&lt;/sub&gt;</td>
<td>.964</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>1MTP&lt;sub&gt;x,y,z&lt;/sub&gt;, MF&lt;sub&gt;x&lt;/sub&gt;, GR&lt;sub&gt;x&lt;/sub&gt;</td>
<td>.965</td>
<td>4</td>
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<td></td>
<td>Full model</td>
<td>.968</td>
<td>4</td>
</tr>
<tr>
<td>2.2 Hz (anteroposterior)</td>
<td>MF&lt;sub&gt;y&lt;/sub&gt;, 1MTP&lt;sub&gt;y&lt;/sub&gt;</td>
<td>.960</td>
<td>11</td>
</tr>
<tr>
<td></td>
<td>MF&lt;sub&gt;y&lt;/sub&gt;, 1MTP&lt;sub&gt;y&lt;/sub&gt;, 5MTP&lt;sub&gt;z&lt;/sub&gt;</td>
<td>.964</td>
<td>10</td>
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<td></td>
<td>Full model</td>
<td>.973</td>
<td>9</td>
</tr>
<tr>
<td>3.0 Hz mediolateral (left)</td>
<td>1MTP&lt;sub&gt;x&lt;/sub&gt; and 3MTP&lt;sub&gt;x&lt;/sub&gt;</td>
<td>.976</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>1MTP&lt;sub&gt;x&lt;/sub&gt;, 3MTP&lt;sub&gt;x&lt;/sub&gt;, 5MTP&lt;sub&gt;z&lt;/sub&gt;</td>
<td>.985</td>
<td>3</td>
</tr>
<tr>
<td></td>
<td>Full model</td>
<td>.987</td>
<td>3</td>
</tr>
<tr>
<td>3.0 Hz mediolateral (right)</td>
<td>1MTP&lt;sub&gt;x&lt;/sub&gt;, 3MTP&lt;sub&gt;x&lt;/sub&gt;, GR&lt;sub&gt;x&lt;/sub&gt;</td>
<td>.988</td>
<td>5</td>
</tr>
<tr>
<td></td>
<td>Full model</td>
<td>.996</td>
<td>3</td>
</tr>
<tr>
<td>3.0 Hz (anteroposterior)</td>
<td>MF&lt;sub&gt;y&lt;/sub&gt; and 3MTP&lt;sub&gt;y&lt;/sub&gt;</td>
<td>.977</td>
<td>9</td>
</tr>
<tr>
<td></td>
<td>MF&lt;sub&gt;y&lt;/sub&gt;, 3MTP&lt;sub&gt;y&lt;/sub&gt;, 5MTP&lt;sub&gt;y&lt;/sub&gt;</td>
<td>.978</td>
<td>9</td>
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<tr>
<td></td>
<td>Full model</td>
<td>.986</td>
<td>7</td>
</tr>
<tr>
<td>Self-selected mediolateral (left)</td>
<td>3MTP&lt;sub&gt;x&lt;/sub&gt;, GR&lt;sub&gt;x&lt;/sub&gt;</td>
<td>.977</td>
<td>8</td>
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<tr>
<td></td>
<td>3MTP&lt;sub&gt;x&lt;/sub&gt;, GR&lt;sub&gt;x&lt;/sub&gt;, 5MTP&lt;sub&gt;z&lt;/sub&gt;</td>
<td>.980</td>
<td>7</td>
</tr>
<tr>
<td></td>
<td>Full model</td>
<td>.983</td>
<td>7</td>
</tr>
<tr>
<td>Self-selected mediolateral (right)</td>
<td>1MTP&lt;sub&gt;x&lt;/sub&gt;, 5MTP&lt;sub&gt;x,z&lt;/sub&gt;</td>
<td>.971</td>
<td>5</td>
</tr>
<tr>
<td></td>
<td>1MTP&lt;sub&gt;x&lt;/sub&gt;, 5MTP&lt;sub&gt;x,y,z&lt;/sub&gt;, 3MTP&lt;sub&gt;x&lt;/sub&gt;</td>
<td>.977</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>Full model</td>
<td>.985</td>
<td>3</td>
</tr>
<tr>
<td>Self-selected (anteroposterior)</td>
<td>MF&lt;sub&gt;y,z&lt;/sub&gt;, GR&lt;sub&gt;y&lt;/sub&gt;</td>
<td>.963</td>
<td>11</td>
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<td></td>
<td>1MTP&lt;sub&gt;x&lt;/sub&gt;, MF&lt;sub&gt;y,z&lt;/sub&gt;, GR&lt;sub&gt;y&lt;/sub&gt;</td>
<td>.965</td>
<td>11</td>
</tr>
<tr>
<td></td>
<td>Full model</td>
<td>.972</td>
<td>10</td>
</tr>
</tbody>
</table>
Table 9. Forced entry regression: equations derived for each frequency in each direction with R, R², standard error of the estimate (SEE) and cross-validation results. Equation is based on marker position data in metres, SEE and cross-validation results are presented in millimetres for clarity.

<table>
<thead>
<tr>
<th>Frequency and direction</th>
<th>Equation</th>
<th>R</th>
<th>R²</th>
<th>Adjusted R²</th>
<th>SEE</th>
<th>Cross-validated RMSD (all)</th>
<th>Mean difference</th>
<th>Standard deviation</th>
<th>95% confidence interval</th>
</tr>
</thead>
<tbody>
<tr>
<td>Quiet standing (left)</td>
<td>CoP_x = 0.059 + 0.138(1MTP_x) + 0.186(1MTP_y) – 1.430(5MTP_z) + 0.681(MF_z) + 0.090(MF_y) - 0.233(MF_z)</td>
<td>1.00</td>
<td>.999</td>
<td>.999</td>
<td>1</td>
<td>6</td>
<td>3</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>Quiet standing (right)</td>
<td>CoP_x = 0.390 + 0.476(1MTP_x) + 0.952(1MTP_y) + 0.200(1MTP_z) – 0.799(5MTP_y) – 0.889(5MTP_z) – 0.114(MF_z) – 1.190(MF_y)</td>
<td>.999</td>
<td>.998</td>
<td>.998</td>
<td>1</td>
<td>4</td>
<td>2</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>Quiet standing (anteroposterior)</td>
<td>CoP_y = -0.237 - 0.203(1MTP_x) + 1.739(1MTP_y) + 0.591(1MTP_z) - 0.392(5MTP_y) + 1.139(5MTP_z) + 3.848(5MTP_y) + 0.785(MF_y) – 1.707(MF_z) – 2.059(MF_y)</td>
<td>.993</td>
<td>.987</td>
<td>.987</td>
<td>3</td>
<td>12</td>
<td>9</td>
<td>-1</td>
<td>9</td>
</tr>
</tbody>
</table>
1.5 Hz
mediolateral
(left)
\[
\text{CoP}_X = -0.014 + 0.941(1\text{MTP}_X) - 0.116(1\text{MTP}_Y) \\
-1.281(1\text{MTP}_Z) - 0.311(5\text{MTP}_X) - 0.034(5\text{MTP}_Y) \\
+ 0.308(5\text{MTP}_Z) + 0.408(\text{MF}_X) + 0.154(\text{MF}_Y) + \\
0.071(\text{MF}_Z)
\]

1.5 Hz
mediolateral
(right)
\[
\text{CoP}_X = 0.070 + 0.080(1\text{MTP}_X) + 0.384(1\text{MTP}_Y) \\
+ 1.015(1\text{MTP}_Z) - 0.332(5\text{MTP}_X) - \\
0.477(5\text{MTP}_Y) + 0.762(5\text{MTP}_Z) + 1.246(\text{MF}_X) + \\
0.099(\text{MF}_Y) - 1.254(\text{MF}_Z)
\]

1.5 Hz
(anteroposterior)
\[
\text{CoP}_Y = 0.017 + 0.007(1\text{MTP}_X) + 0.495(1\text{MTP}_Y) \\
- 1.571(1\text{MTP}_Z) + 0.072(5\text{MTP}_X) + \\
0.056(5\text{MTP}_Y) + 0.691(5\text{MTP}_Z) - 0.058(\text{MF}_X) + \\
0.452(\text{MF}_Y) + 0.444(\text{MF}_Z)
\]

2.2 Hz
mediolateral
(left)
\[
\text{CoP}_X = 0.050 + 0.537(1\text{MTP}_X) - 0.556(1\text{MTP}_Y) \\
- 0.609(1\text{MTP}_Z) - 0.218(5\text{MTP}_X) - \\
0.093(5\text{MTP}_Y) + 0.099(5\text{MTP}_Z) + 0.592(\text{MF}_X) + \\
0.671(\text{MF}_Y) - 0.053(\text{MF}_Z)
\]
<table>
<thead>
<tr>
<th>Frequency</th>
<th>Configuration</th>
<th>CoP X</th>
<th>CoP Y</th>
<th>CoP Z</th>
<th>1MTP X</th>
<th>1MTP Y</th>
<th>1MTP Z</th>
<th>5MTP X</th>
<th>5MTP Y</th>
<th>5MTP Z</th>
<th>MF X</th>
<th>MF Y</th>
<th>MF Z</th>
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</thead>
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<tr>
<td>2.2 Hz</td>
<td>mediolateral</td>
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<td>0.051</td>
<td>0.390</td>
<td>0.239</td>
<td>0.448</td>
<td>-0.235</td>
<td>0.192</td>
<td>+0.391</td>
<td>+0.861</td>
<td>-0.021</td>
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<tr>
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<td>(right)</td>
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</tr>
<tr>
<td></td>
<td></td>
<td>.983</td>
<td>.966</td>
<td>.966</td>
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<td>0</td>
<td>4</td>
<td>8</td>
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<td></td>
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<td>2.2 Hz</td>
<td>anteroposterior</td>
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<td></td>
<td></td>
<td>0.025</td>
<td>0.277</td>
<td>0.516</td>
<td>-1.851</td>
<td>0.075</td>
<td>0.087</td>
<td>+0.617</td>
<td>-0.303</td>
<td>+0.445</td>
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<td>(anteroposterior)</td>
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<tr>
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<td>.984</td>
<td>.969</td>
<td>.968</td>
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<td>3.0 Hz</td>
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<td>0.001</td>
<td>0.566</td>
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<td>-0.859</td>
<td>0.205</td>
<td>0.216</td>
<td>+0.356</td>
<td>+0.643</td>
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</tr>
<tr>
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<td>(left)</td>
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<td>.988</td>
<td>.988</td>
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<td>5</td>
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<tr>
<td>3.0 Hz</td>
<td>mediolateral</td>
<td></td>
<td></td>
<td></td>
<td>0.024</td>
<td>0.367</td>
<td>0.178</td>
<td>+0.872</td>
<td>-0.311</td>
<td>0.031</td>
<td>+0.343</td>
<td>+0.976</td>
<td>-0.204</td>
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<tr>
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<td>(right)</td>
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<td></td>
<td></td>
<td>.997</td>
<td>.994</td>
<td>.994</td>
<td>3</td>
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<td>0</td>
<td>3</td>
<td>6</td>
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<td></td>
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<tr>
<td></td>
<td>CoP&lt;sub&gt;y&lt;/sub&gt; = 0.091 + 0.209(1MTP&lt;sub&gt;x&lt;/sub&gt;) + 0.545(1MTP&lt;sub&gt;y&lt;/sub&gt;)</td>
<td>CoP&lt;sub&gt;y&lt;/sub&gt; = 0.025 + 0.617(1MTP&lt;sub&gt;x&lt;/sub&gt;) – 0.319(1MTP&lt;sub&gt;y&lt;/sub&gt;)</td>
<td>CoP&lt;sub&gt;y&lt;/sub&gt; = 0.036 + 0.173(1MTP&lt;sub&gt;x&lt;/sub&gt;) + 0.340(1MTP&lt;sub&gt;y&lt;/sub&gt;)</td>
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<tr>
<td></td>
<td>– 2.216(1MTP&lt;sub&gt;z&lt;/sub&gt;) + 0.339(5MTP&lt;sub&gt;x&lt;/sub&gt;) + 0.259(5MTP&lt;sub&gt;y&lt;/sub&gt;) + 0.866(5MTP&lt;sub&gt;z&lt;/sub&gt;) – 0.567(MF&lt;sub&gt;x&lt;/sub&gt;) + 0.227(MF&lt;sub&gt;y&lt;/sub&gt;) – 0.110(MF&lt;sub&gt;z&lt;/sub&gt;)</td>
<td>– 0.994(1MTP&lt;sub&gt;z&lt;/sub&gt;) – 0.202(5MTP&lt;sub&gt;x&lt;/sub&gt;) + 0.258(5MTP&lt;sub&gt;y&lt;/sub&gt;) + 0.612(5MTP&lt;sub&gt;z&lt;/sub&gt;) + 0.561(MF&lt;sub&gt;x&lt;/sub&gt;) + 0.065(MF&lt;sub&gt;y&lt;/sub&gt;) – 0.042(MF&lt;sub&gt;z&lt;/sub&gt;)</td>
<td>– 1.470(1MTP&lt;sub&gt;z&lt;/sub&gt;) + 0.055(5MTP&lt;sub&gt;x&lt;/sub&gt;) + 0.032(5MTP&lt;sub&gt;y&lt;/sub&gt;) + 0.594(5MTP&lt;sub&gt;z&lt;/sub&gt;) – 0.193(MF&lt;sub&gt;x&lt;/sub&gt;) + 0.663(MF&lt;sub&gt;y&lt;/sub&gt;) + 0.095(MF&lt;sub&gt;z&lt;/sub&gt;)</td>
<td></td>
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<td></td>
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</tr>
<tr>
<td>(anteroposterior)</td>
<td>.992 .984 .984 8 10 8 -1 8 16</td>
<td>.991 .983 .983 7 4 4 1 4 7</td>
<td>.982 .965 .965 11 11 11 2 11 21</td>
<td></td>
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</tr>
</tbody>
</table>
6.4.4 Effect on joint torque calculation

During hopping at 1.5 Hz, 2.2 Hz, 3.0 and the SS frequency, 106, 180, 232 and 201 hops respectively were used for analysis. Similar maximum moments were observed across all frequencies, with calculated moments of 142 ±46 N.m (1.5 Hz), 135 ±65 N.m (2.2 Hz), 137 ±33 N.m (3.0 Hz) and 140 ±70 N.m (SS) using the CoP from the FP. The corresponding moments using the regression method were 135 ±38 N.m, 145 ±72 N.m, 148 ±40 N.m and 148 ±72 N.m. When horizontal force was removed, these values were 134 ±39 N.m, 145 ±72 N.m, 147 ±40 N.m and 146 ±70 N.m. The maximum moments using the regression method were 139 ±41 N.m, 149 ±77 N.m, 148 ±40 N.m and 150 ±73 N.m. These values slightly changed with removal of the horizontal force to 137 ±40 N.m, 149 ±77 N.m, 147 ±40 N.m and 148 ±70 N.m. Differences in magnitude of maximum and corresponding moments between the three methods are presented in Table 10. The RMSD between methods was similar across frequencies.

A mean difference in the timing of maximum moment of 3 ±90 ms, 2 ±9 ms, 0 ±3 ms and 0 ±16 ms was observed using the regression equations, which changed when horizontal force was removed to -2 ±83 ms, 3 ±10 ms, 0 ±4 ms and 1 ±15 ms. When these were considered as absolute differences, timing differences were 36 ±83 ms, 5 ±8 ms, 1 ±3 ms and 7 ±14 ms. When horizontal force was removed, these differences were 30 ±77 ms, 5 ±9 ms, 1 ±3 ms and 7 ±14 ms.

6.5 Discussion

Table 6 clearly shows the error in assumption of a fixed centre of pressure, with a large difference in the distance the CoP moved between 1.5 Hz and 3.0 Hz. While ML movement was similar, the range of AP movement reduced by almost half. This is probably because the decreased CT during hopping at higher frequencies does not allow for large ranges of movement in a short period of time. These results support the need for a method which can determine dynamic CoP movement during a hopping task.

It is not surprising that marker position in the axis of interest is the marker kinematic most related to CoP position as similar measures were compared. Due to the high predictive abilities of the initial stepwise regressions, the addition of extra markers did not improve
<table>
<thead>
<tr>
<th>Actual hopping frequency (Hz)</th>
<th>Force input</th>
<th>RMSD (N.m)</th>
<th>RMSD (percent of maximum: %)</th>
<th>FP maximum-correcting (absolute: N.m)</th>
<th>FP maximum-correcting (percent: %)</th>
<th>Difference in maximum moment (absolute: N.m)</th>
<th>Difference in maximum moment (percent: %)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 Hz 1.51 ±0.07</td>
<td>Horizontal force</td>
<td>8 ±6</td>
<td>5.6 ±2.9</td>
<td>10 ±10</td>
<td>6.6 ±4.7</td>
<td>10 ±11</td>
<td>6.6 ±6.0</td>
</tr>
<tr>
<td></td>
<td>No horizontal force</td>
<td>8 ±6</td>
<td>5.5 ±2.8</td>
<td>10 ±10</td>
<td>6.2 ±4.8</td>
<td>9 ±10</td>
<td>6.0 ±5.2</td>
</tr>
<tr>
<td>2.2 Hz 2.21 ±0.07</td>
<td>Horizontal force</td>
<td>7 ±6</td>
<td>5.1 ±3.4</td>
<td>12 ±16</td>
<td>8.6 ±7.9</td>
<td>16 ±24</td>
<td>10.6 ±12.5</td>
</tr>
<tr>
<td></td>
<td>No horizontal force</td>
<td>7 ±6</td>
<td>5.4 ±3.3</td>
<td>13 ±16</td>
<td>8.8 ±8.0</td>
<td>16 ±24</td>
<td>10.8 ±12.5</td>
</tr>
<tr>
<td>3.0 Hz 2.98 ±0.12</td>
<td>Horizontal force</td>
<td>6 ±4</td>
<td>4.4 ±2.3</td>
<td>12 ±9</td>
<td>7.8 ±5.0</td>
<td>12 ±9</td>
<td>7.9 ±5.0</td>
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<tr>
<td></td>
<td>No horizontal force</td>
<td>6 ±4</td>
<td>4.5 ±2.1</td>
<td>11 ±8</td>
<td>7.4 ±5.0</td>
<td>11 ±8</td>
<td>7.4 ±4.9</td>
</tr>
<tr>
<td>SS 2.28 ±0.18</td>
<td>Horizontal force</td>
<td>7 ±5</td>
<td>6.0 ±4.2</td>
<td>13 ±13</td>
<td>9.7 ±7.9</td>
<td>15 ±17</td>
<td>11.8 ±11.3</td>
</tr>
<tr>
<td></td>
<td>No horizontal force</td>
<td>8 ±4</td>
<td>6.2 ±4.0</td>
<td>12 ±11</td>
<td>9.0 ±7.5</td>
<td>14 ±14</td>
<td>10.7 ±10.3</td>
</tr>
</tbody>
</table>
predictive power of the equations by any practically significant amount. This led to the conclusion that use of large numbers of passive markers in close proximity to each other (which may make tracking difficult) is unnecessary and not justified. The X and Y axis feature heavily as predictors for ML CoP. The Y axis features heavily in AP CoP determination as does the Z axis, due to its relationship with the angle the foot is acting upon the ground. Increased Z during ground contact with similar X and Y location suggests the foot is positioned at a steeper angle from the horizontal, which results in the CoP acting further forwards in the anterior direction than when Z is lower (decreased foot angle). The MF marker appears frequently, probably as it is located close to the centre of mass of the foot.

For standing and all frequencies, CoP predictive ability was greater in the ML direction than AP. FP accuracy is greater in the short (X) axis than the long (Y) axis due to the smaller distance between the transducers used to determine the forces and position (Bobbert and Schamhardt 1990; Middleton et al. 1999). CoP movement in the ML direction is also smaller than the AP direction during unidirectional, single plane hopping therefore the predictive equation only needs to operate within a small range.

Use of a reduced number of markers resulted in similar SEE to that of a full five marker set (Table 9). Considering the accuracy of this FP for CoP is 1 to 1.5 mm in static loading, plate noise is 0.4 to 0.7 mm, re-zeroing in post-processing may affect CoP by a further 1 to 3 mm and previously reported inaccuracies of measurement of CoP using the FP during dynamic tasks (Bobbert and Schamhardt 1990), it appears the motion analysis system performed well in determining CoP. SEE results suggest good predictive abilities of the regression equations with maximum ML SEE of 8 mm and AP SEE of 12 mm for all frequencies apart from 1.5 Hz. RMSD for the cross-validated participants for most conditions were similar to SEE. Cross-validated RMSD was lower than those previously reported. Pillet et al. (2010) reported RMSD of approximately 18 mm (ML) and 43 mm (AP) during walking. This method may be more accurate because the markers used are closer to the CoP than the full body marker sets used in their study. Removal of spatial and temporal error may explain the lower differences in CoP position and timing of maximum moment reported here compared to Chesnin et al. (2000), Fradet et al. (2009) and Chumanov et al. (2010). Mean difference in CoP position was low with correspondingly
low 95% CI, and very large linear relationships between CoP using both methods were observed.

Cross-validated RMSD in 1.5 Hz hopping was 14 mm (ML) and 20 mm (AP). There were differences between SEE and cross-validated RMSD of 1 to 1.1 mm (ML) and 8 mm (AP). The 95% CI for difference between methods was very large (28 mm), and the timing of maximum moment using the two methods was excessively large with a mean absolute difference of 36 ±83 ms. This suggests the regression method is inappropriate for 1.5 Hz hopping, probably due to the increased variability and range of motion of the ankle at this particular frequency with corresponding decreased joint stiffness (Hobara et al. 2011), excessive foot spread during impact and presence of double-peak vertical ground reaction forces.

The differences in calculated moments are slightly greater than the 5 N.m RMSD reported by Forner-Cordero et al. (2006) for sagittal plane ankle torque during a walking task, and similar to the 6.91 N.m RMSD of Fong et al. (2008) for ankle supination torque during a number of dynamic tasks. However, neither of these papers reported the magnitude of difference between calculated maximum moment so may be negatively affected by very low difference prior to or after the maximum, nor do they report differences in the timing of the maximum moment. Review of their graphs would suggest that their proposed methods, despite slightly lower RMSD, resulted in altered timing of varying amount of the maximum moment. This variable provides further evidence that this marker based method is unsuitable for analysis of 1.5 Hz hopping. Even though similar RMSD in moment was found and the difference in maximum moment was small (<7%), very large timing differences were observed. Mean difference was -2 ±83 ms with absolute difference of 30 ±77 ms, which is of particular importance when a variable such as joint stiffness or rate of loading is of interest. It is difficult to compare this timing to that observed in other methods, as no other author appears to have reported it. A timing error of this magnitude is excessively large to consider using the regression method at this frequency when these measures are of interest.

One source of this difference is probably the large difference in CoP position observed during the cross-validation. It may also be due to an altered kinematic strategy used by the limb during 1.5 Hz hopping to maintain the low hopping frequency. This may not only affect the CoP calculation, but also the kinematic input into the inverse dynamics analysis.
In contrast, the linear behaviour at higher frequencies and more consistent kinematics may explain the improved accuracy during 2.2 Hz, 3.0 Hz and SS hopping.

The removal of the horizontal force did not have any significant effect on the RMSD or magnitude of the maximum moment during an impact. Even though horizontal forces of approximately 60 N were observed, it is not unexpected that there was little effect of its removal on the resultant moments as the vertical force has a much larger influence on the moment itself, and its associated moment arm is small. Its moment is subsequently small so its removal has little influence on the calculated moment.

6.6 Direction of future research in the context of the thesis

The results of this study show a marker-based method of determining CoP can be used during a two-legged hopping task at frequencies greater than 1.5 Hz, with maximum accuracy observed during short contact hopping at 3.0 Hz. The preceding chapters have shown that the sledge can be used to quantify plantarflexor function during a dynamic SSC task, but more in-depth analysis of joint kinetics and MTU mechanics would provide further insight into our understanding of MTU behaviour during this task. Chapter 7 combines the adapted force sledge and marker method of determining CoP with ultrasound, to investigate the main factors influencing stiffness during a SSC task, and potential surrogate measures of Achilles stiffness and loading.
Chapter 7: Assessment of plantarflexor mechanical behaviour during a stretch-shortening cycle task
7.1 Abstract

Assessment of the plantarflexors during a stretch-shortening cycle task is important, due to their role in locomotion and high rate of injury. Stiffness is a measure of particular interest, due to its reported link with performance and injury. Due to methodological problems, it has been difficult in the past to establish how measures of muscle function relate to stiffness modulation. Surrogate measurement of underlying muscle-tendon unit function is also of interest, especially for the practitioner in the monitoring of muscle and tendon health. The aims of this chapter were to identify the relationship between measures of muscle-tendon unit function and plantarflexor stiffness during a fast stretch-shortening cycle task and to investigate if force development time, elastic charge time or reactive strength index were representative of muscle-tendon unit behaviour. The rate of peak force development, rate of force development in the first 30 ms and contact time were found to be important determinants of plantarflexor stiffness. Force development time, elastic charge time and reactive strength index were not representative of tendon stiffness or tendon loading, but reactive strength index was related to the amount of energy absorbed and generated during loading. These results highlight the link between muscle function and modulation of stiffness, as well as the difficulties in surrogate measurement of tendon behaviour during dynamic activities.
7.2 Introduction

The plantarflexors have an important role to play in locomotion and stability, and have been shown to be the primary energy absorbers and generators during both acceleration and maximal velocity sprinting (Johnson and Buckley 2001; Bezodis et al. 2008). Their distal location and importance in movement has subsequently led to this muscle group and the ankle joint being one of the most commonly injured areas of the lower limb (Fong et al. 2007). Assessment and understanding of their function in dynamic tasks is of interest to both researchers and practitioners, to enhance athletic performance as well as guide rehabilitation. Stiffness is a mechanical property which has attracted much interest in recent years due to reported links to enhanced performance and injury. As the ratio between applied force to linear or angular deformation (Zatsiorsky and Prilutsky 2012), it can be evaluated at whole limb, segment, joint or tissue level.

Leg stiffness is a quasi-stiffness which reflects the stiffness of all the joints and component tissues (such as muscle, tendon and ligament) of the lower limb (Latash and Zatsiorsky 1993). Stiffness at joint level is modulated by the stiffness of its intrinsic structures, as well as the level of muscle activation. Joint stiffness has previously been shown to vary with task demands, with the knee shown to be important during maximal effort tasks and the ankle more important during high frequency loading tasks (Hobara et al. 2009; Hobara et al. 2011). While the stiffness of the muscle and tendon are clearly important to modulation of overall stiffness, the relative contribution of musculotendinous mechanical behaviour and kinetic-derived parameters to overall stiffness is not fully understood. For example, power trained athletes have been shown to demonstrate increased stiffness of the leg and ankle joint during cyclical loading (Harrison et al. 2004; Hobara et al. 2008) but at tendon level, tend to demonstrate similar Achilles tendon stiffness as controls (Kubo et al. 2000a). This suggests the increase in ankle stiffness may come primarily from the muscle and muscle related factors such as changes in rate of force development of the muscle.

Understanding the contributing factors to ankle stiffness is challenging as most methods of assessing plantarflexor function assess the isolated muscle during isometric or isokinetic contractions rather than a SSC condition. Recent work by Furlong and Harrison (2013) used an adapted force sledge apparatus to isolate the plantarflexors for analysis during a dynamic, controlled fast SSC task. This method uses similar kinematics as those observed during hopping and running tasks, with comparable contact times. This method allows for
assessment of joint stiffness during a fast SSC task while simultaneously quantifying rates of force development and joint kinetics.

One of the factors thought to have a significant influence on ankle joint stiffness is the proximity of the large Achilles tendon to the joint. Measurement of the mechanical properties of tendon is a complex procedure requiring expensive equipment and a high level of technical expertise. The most common method of determining these properties combine a method of measuring joint torque or force with ultrasound imaging. Isometric dynamometry is most commonly used to determine ankle torque and subsequently tendon force (Kubo et al. 2000a; Arampatzis et al. 2006) but Lichtwark and Wilson (2005b) combined motion analysis and force plates in a method with high ecological validity and portability which can be used during a hopping task. However, neither of these methods are practical for use by clinicians or coaches who may want to obtain large amounts of data, quickly and easily, in a short space of time.

The electromechanical delay (EMD) of a muscle refers to the time difference between the onset of muscle activation and the first registration of force. It is measured on a muscle-specific basis, and can be obtained during voluntary or involuntary (twitch) contractions. It is still not clear what the electromechanical delay actually measures, but it is highly likely that it is dependent on a combination of biochemical and mechanical factors. Cavanagh and Komi (1979) suggested EMD is a measure of the time taken to stretch the series elastic component of the muscle-tendon unit i.e. a comparable measure to stiffness. Similarly, Waugh et al. (2013) found Achilles tendon stiffness explained 68% of the variance in measures of voluntary EMD of the gastrocnemius, suggesting the mechanical properties of the tendon make a large contribution to the EMD. Kubo et al. (2005b) suggested the measure may be more related to biochemical rather than mechanical factors as they observed large changes in EMD but no difference in the stiffness of the tendon itself with heating and cooling of the limb. As the definition of EMD is highly dependent on timing of muscle activation, this would suggest a large involvement of biochemical processes. Winter and Brookes (1991) proposed an expanded definition of EMD where the initial difference between onset of activation and force was termed force development time (FDT) and the time difference between registration of force and raising of the heel (as determined by release of a heel switch) was termed elastic charge time (ECT). Based on the rationale above, it is possible that the ECT portion of the EMD may be a more related
measure of the stiffness of the tendinous structures. If so, ECT could provide a simple, easy, quick method for the practitioner or coach to determine the stiffness of the tendon.

Similarly, the reactive strength index (RSI) has been considered indicative of muscle-tendon unit (MTU) mechanical behaviour as increased RSI has been observed in sprint and power athletes (Harrison et al. 2004). RSI is defined as the ratio between height jumped and its preceding contact time. It indicates the ability of a participant to effectively utilise the SSC (Young 1995) or their ‘explosiveness’ (Flanagan et al. 2008). Improved SSC effectiveness is usually attributed to increased MTU stiffness. In recent years, RSI has grown in popularity as a quick method for the strength and conditioning professional to assess an athlete’s ability to effectively utilise eccentric contractions to maximise concentric phase performance. It has also been proposed for use in the monitoring of fast SSC training effectiveness (Flanagan and Comyns 2008). Despite frequent reference to RSI as a method of quantifying training intensity and musculoskeletal loading (McClymont 1995; McMahon et al. 2012) and to indicate the strain on the MTU during a SSC task (Lloyd et al. 2009), there is little literature to justify this. There is also little literature simultaneously quantifying RSI, joint kinetics or rates of force development during plyometric tasks. While Jensen and Ebben (2007) reported eccentric rates of force development and knee joint kinetics during a number of plyometric exercises, no RSI values were reported. The relationship between muscle function, loading and RSI is hence still unclear.

Stiffness is of particular interest to the researcher and practitioner due to reported links with performance and injury. Our understanding of the relationship between muscle function and mechanics, and the factors and processes determining stiffness are still quite limited. The aim of this study was to identify the main predictors of plantarflexor stiffness $K_{PF}$ during this low-load plyometric task. A secondary aim was to determine the relationship between ECT, FDT, RSI and MTU behaviour. The results of this study have implications for our understanding of the processes underlying stiffness and stiffness modulation as well as for assessment of plantarflexor loading and mechanical behaviour.

### 7.3 Methods

#### 7.3.1 Participants

Following institutional ethical approval, 19 healthy adults gave written informed consent to participate in this study (8 males, 11 females, age: 25.2 ±2.35 years, mass: 74.0 ±11.0,
height: 1.73 ±0.10 m). All were active a minimum of three days per week for 30 minutes per day, and capable of participating in exercise requiring a number of impacts. None had a history of lower limb surgery and all were injury free for the preceding 3 months. Participants were requested to refrain from unaccustomed strenuous exercise for the 24 hours preceding testing.

7.3.2 Participant preparation and anthropometrics

Prior to commencement of testing, 9.5 mm retro-reflective markers were attached to the superior 1st metatarsophalangeal joint (1MTP), lateral 5th metatarsophalangeal joint (5MTP), a point in line with the 3rd metatarsophalangeal joint at half the length of the foot (MF), the lateral malleolus (MALL), calcaneus (CALC) and estimated knee joint centre (KJC). All measures were obtained using the limb the participant would preferentially hop on. For 17 participants this was the right leg and for two participants, it was the left.

Foot length was measured using a custom-built measuring board with foot length defined as the sagittal plane distance between the posterior calcaneus and the longest toe. Pilot work has shown the reliability of this measure to be very high (single ICC = 0.95, average ICC = 0.99, standard deviation = 0.1 cm, coefficient of variation = 0.31%) so one measure was obtained.

To obtain images of the muscle-tendon junction, a 128-element, linear multi-frequency 7.5MHz veterinary rectal ultrasound probe with a 65 mm field-of-view was positioned over the muscle-tendon junction (MTJ) of distal gastrocnemius and the Achilles tendon. It was secured in position with elasticated self-adhesive bandage (Powerflex, Andover Healthcare, Salisbury, MA., USA). Prior to final positioning, the researcher checked that the MTJ did not move out of the field of view during the test task. A PC-based ultrasound system (Echoblaster 128, Telemed, Vilnius, Lithuania) was used to image the junction at 50 Hz. Four motion analysis markers were attached to this probe using a specially made cover to allow for the identification of the probe and its local coordinate system in the global coordinate system (Figure 10).
7.3.3 Force sledge protocol

An adapted force sledge apparatus consisting of a wooden plate free to move along rails angled at 30° to the horizontal was used for this study. Another 9.5 mm marker was placed on the side of this plate for tracking by three-dimensional motion analysis (200 Hz, MAC Eagle, Motion Analysis Corporation Inc., Santa Rosa, CA., USA). Participants were positioned supine at the base of the sledge with the thigh secured to a solid support at its proximal and distal ends. Hip angle was approximately 135° and knee angle has been shown previously to remain between 140° and 160° during impact. The ankle remained free to push the plate up the inclined rails (Figure 11).
Once secured at the base of the sledge, participants were instructed to strike the plate rhythmically in a marked area pushing it as high and as fast as they could using only the ankle joint. They were initially asked to push the plate away from them using the above criteria then the plate was dropped towards their foot from a 30 cm height after a ‘3, 2, 1’ countdown. This protocol was repeated for a minimum of 25-30 impacts until the participant was satisfied that they were familiar with the task and the researcher judged them to be striking the plate as instructed.

The pre-conditioning protocol commonly used with isometric test conditions (Maganaris 2003b; Pearson et al. 2007) was not used due to the greatly decreased duration of loading on the tendon during each individual contraction (<0.25 s) compared to the typical isometric protocol (3-10 s). Farris et al. (2012) reported no change in stiffness, strain or tendon length after a 30 minute run which suggests a lack of creep in SSC tasks, and that creep only needs to be considered during contractions of long duration.

Similar to familiarisation, the plate began at a position 30 cm above the foot and was released after a ‘3, 2, 1’ countdown. Participants were instructed to continue striking the plate for a total of 11 impacts. Following each trial, the plate was secured away from the foot. A similar protocol to that of a repetition maximum test was used, with the ultimate aim to achieve the maximum loading a participant could exert for 11 impacts. Participants were allowed at least two minutes rest in between trials but if more time was required this was allowed. A loading equivalent to 70% of this 11RM (actual loading used: 30.1 ±3.8 kg,
69.9 ±1.1% of 11RM, 41.4 ±7.7% of body mass) was used for the test trial as this has been shown to be reliable and a loading that participants were comfortable with completing multiple impacts.

7.3.4 Sledge-derived variable definition and calculation

Markers were filtered using a fourth order, zero lag, low-pass Butterworth filter. Residual analysis was conducted to identify the optimum cut-off frequency to ensure an optimal signal: noise ratio (Winter 2005), with a cut-off frequency of 12 Hz used for the sledge marker. A cut-off of 13 Hz was used for the 5MTP marker, 14 Hz for the 1MTP, MALL, CALC and probe markers, 17 Hz for the mid-foot marker and 18 Hz for the knee marker. Filtering was completed in Motion Analysis software (Cortex v3.1.1.1290, Motion Analysis Corporation Inc., Santa Rosa, CA., USA). All variables were calculated in Microsoft Excel (Microsoft Inc., Redmond, WA., USA) with the 3rd to 5th impacts chosen for analysis as previous work found middle impacts to be the most reliable for analysis. These impacts were used instead of the 5th to 7th as used in other chapters due to technical difficulties with the graphics card used for data collection, which resulted in freezing of the ultrasound images in the latter portion of trials.

Plate acceleration was calculated as the second derivative of plate position with respect to time, with force calculated using Newton’s second law with a correction for the component of weight acting down the sledge rails as the sledge was angled at 30°. Contact time (CT) was defined as the period when plate marker acceleration was greater than zero and flight time (FT) defined as the period when it was zero or less. These definitions were used as when the sledge was away from the foot it was in free-fall therefore the only forces acting on it were gravity (-4.905 m.s⁻²) and friction. Friction during free-fall has been calculated at 0.18% so was considered negligible for these calculations. Peak force (FP) was the maximum force developed during each contact with rate of peak force development (RPFD) calculated as the peak force divided by the time in seconds it took to reach it. Rate of force development in the first 30 and 50 ms was calculated as the force at 30 and 50 ms, divided by 0.03 and 0.05 s respectively.

Plate height was calculated from the motion analysis system as the difference between the height of the plate at the end of CT to the peak height achieved in the subsequent FT.
Reactive strength index (RSI) was defined as the ratio between plate height and preceding CT.

Plantarflexor stiffness (K_{PF}) was calculated as the ratio between F_p and peak displacement of the sledge in the direction of the sledge rails from the start of CT in metres. Sledge power was calculated as the product of sledge force and sledge velocity. Peak eccentric power (PS_{ECC}) was defined as the minimum power observed during an impact, with peak concentric power (PS_{CONC}) defined as the maximum positive power observed during an impact.

7.3.5 Inverse dynamics analysis to determine ankle joint torque

The centre of pressure (point of application of the sledge reaction force) was calculated using the marker-based method presented in chapter 6. The frequency of sledge impact has been calculated at approximately 0.5-0.8 Hz but due to the similarity of the contact time and shape of the force curve observed during this task to that observed in 3.0 Hz hopping, the 3.0 Hz equation was used for calculating CoP. When considered in conjunction with high-speed video footage, the predicted CoP positions appeared realistic. Ankle joint moment was calculated using standard inverse dynamics equations (Winter 2005). Joint stiffness (K_{ANK}) was defined as the ratio of maximum joint moment to change in joint angle (Kuitunen et al. 2002b), with joint power calculated as the product of joint moment and ankle angular velocity in radians (Bezodis et al. 2008). Peak eccentric ankle joint power (PA_{ECC}) was defined as the maximum negative power observed during an impact, with peak concentric ankle joint power (PA_{CONC}) defined as the maximum positive power observed during an impact.

The contribution of the weight of the foot was not included in moment calculations, as pilot analysis showed the relationship between foot mass and foot length was non-linear and inconsistent. Foot mass is typically calculated using the model of Dempster (1955), Plagenhoef et al. (1983) or de Leva (1996), which assumes foot mass to be a percentage of body mass (between 1.29% and 1.45% of body mass). There was small variability in the size of feet of participants (CoV: 5.8%) but quite large variability in body mass (16.1%). The feet are primarily composed of bony and tendinous structures, therefore it is reasonable to assume foot mass has more of a relationship with foot length than body mass. The difference in variability of measures suggests that use of a fixed percentage of
body mass to determine foot mass is inappropriate for this analysis and may introduce error into calculations. The moment arm between the malleolus and the centre of mass of the foot is also small (most likely less than 10 cm) hence the calculated value itself may be quite small. It was felt that it was better to leave a potentially erroneous mass calculation out of the calculation rather than include it.

As the influence of body mass and muscle size on force output is well established, force, force development, $K_{PF}$ and ankle kinetics were normalised to total plantarflexor muscle volume by dividing observed forces by personalised total plantarflexor muscle volume. Muscle volume was calculated for each participant using the simple regression equations of Handsfield et al. (2014) for soleus, gastrocnemius medialis and lateralis, peroneus longus and brevis, tibialis posterior, flexor hallucis longus and flexor digitorum longus based on height and mass products. Total plantarflexor muscle volume was calculated as the sum of the combined individual muscle volumes.

7.3.6 Calculation of tendon forces, elongation and stiffness

Tendon force was calculated by dividing calculated ankle moment by the Achilles tendon moment arm, which was defined as the perpendicular distance between the malleolus and the line of action of the Achilles tendon (Lichtwark and Wilson 2005b). An image-based method of determining individualised tendon moment arm similar to Dixon and Kerwin (1998) was used, as the size of the Achilles moment arm is known to vary between individuals. Participants were recorded striking the sledge plate using a high-speed camera operating at 300 Hz (Casio Exelim EX-F1, Tokyo, Japan). The camera was positioned perpendicular to the plane of interest, approximately 1 m from the foot. A 20 cm marked distance on the edge of the sledge plate was used to calibrate the video image. It was assumed all force in the Achilles was generated by the plantarflexor muscles (Lichtwark and Wilson 2005b). The distance between the lateral malleolus and the line of action of the Achilles tendon was digitised manually using SiliconCoach (SiliconCoach Pro, SiliconCoach, Dunedin, New Zealand). Previous work has shown variation in the size of the moment arm when the foot is in the fully plantarflexed position, but CT in this protocol ended prior to this event. The moment arm was reasonably consistent within an individual in the range of motion observed during the task (mean standard deviation across participants: 2 mm) so the mean digitised distance across the CT for each individual was used for calculation of tendon force throughout the range of motion.
All ultrasound images were digitised in Telemed EchoWave software (v3.57, Telemed, Vilnius, Lithuania), which allows for automatic calculation of length of a line within the image. The accuracy of this measurement was checked by checking that a horizontal line in the image corresponded to the reported size of the field of view of the probe (65 mm). The same value was reported for the length of the field of view of the probe by the motion analysis system, as two markers were also positioned at the edge of the field of view. The muscle tendon junction was defined as the intersection point between the internal deep and superficial aponeurosis. Tendon length was calculated using simple trigonometry. The marker on the calcaneus was considered the best approximation for the insertion of the Achilles tendon. Wand markers were positioned at a distance of 35 mm from the middle of the probe. Pythagoras theorem was used to determine the distance between the top of the probe in the plane of the image and the calcaneal marker. The muscle-tendon junction lies at a point below the level of the skin (where the probe is positioned). By using the line of action of the tendon to determine the depth of the tendon, tendon length and length change could also be calculated using Pythagoras theorem. Tendon stiffness ($K_{AT}$) was calculated as the peak tendon force divided by tendon elongation from start of CT to time of peak tendon force. The point where stiffness is measured is known to influence observed stiffness values (Pearson and Onambélé 2012), so it is difficult to compare values reported here with those of other researchers who may have calculated stiffness from 50-100% or 50-80% of the force-elongation curve. Due to technical problems, a total of 27 trials were used for this variable.

### 7.3.7 Electromechanical delay protocol and analysis

A similar method to Winter and Brookes (1990) was used to determine the electromechanical delay of the soleus and medial gastrocnemius muscles. Force development time (FDT) is known as electromechanical delay by other authors (Kubo et al. 2005b; Yavuz et al. 2010; Rampichini et al. 2013; Waugh et al. 2013) and was defined as the time between the onset of muscle activation and development of force. Elastic charge time (ECT) was defined as the time between the onset of force and the movement of the heel from a heel-switch (Figure 12).

Muscle activation was determined using electromyography (EMG). The muscles of interest were soleus and medial gastrocnemius. Electrodes were placed in accordance with SENIAM guidelines. The leg was prepared by shaving and cleaning with alcohol wipes.
The electrodes for soleus were placed at a point 2/3 of the distance from the proximal end of the leg. The electrodes for gastrocnemius medialis were placed on the bulge of the muscle, in the direction of the muscle fibres. The earth electrode was placed on the lateral malleolus for most participants, but due to problems getting the electrode to stick to the malleolus, for three participants it was placed on the tibial plateau.

Participants sat with their knee at 90° and rested quietly with the ball of their foot resting on the force transducer and their heel resting over the heel-switch. To minimise error from soft tissue movement, a square of hardened plastic was attached to the heel. All data was sampled at 1000 Hz (Powerlab, Analog Devices, Limerick, Ireland). Participants were instructed to lift their heel from the heel-switch as quickly as possible after a ‘3, 2, 1’ countdown. A total of eight trials were obtained per participant, with the mean value used for analysis. Trials were eliminated from analysis when the force was observed to either dip prior to registration of force or ramp slowly.
Force development time was defined as the time difference between onset of muscle activation and first registration of force. All muscle activation data was filtered using a band pass filter (between 10 and 500 Hz) and full wave rectified. Onset of muscle activation was determined by visual inspection of a graph of muscle activation versus time. Pilot work showed this to be the most consistent method of determining onset, as false onsets were often detected when objective criteria such as an increase of 2 or 3 standard deviations from baseline were used. The muscle was considered activated at the first time point when a visible, consistent increase in the EMG trace of greater than 2 μV to above 4 μV was observed. Onset of force was defined as the time point where the slope of the force curve was greater than 1. Release of the heel-switch was the time point when voltage rapidly increased from 0 V.

7.3.8 Statistical analysis

All statistical analysis was completed using SPSS Statistics 20 (IBM, Armonk, NY, USA). As this study presents a number of variables not reported previously in this thesis, reliability of measures were established using standard deviation, coefficient of variation, Cronbach’s alpha and two way random intra-class correlation coefficients with absolute agreement. Normality of data was determined using the Shapiro-Wilk’s test. For all statistical tests, alpha was set at $p < 0.05$.

To determine the factors with the greatest influence on $K_{PF}$, multivariate backwards stepwise linear regression was performed. Due to the high observed reliability and recommendation for high $n$ numbers in regression analyses, all three impacts analysed per participant were used for analysis ($n = 57$). Selection of input variables was determined through review of relevant literature, inclusion of variables that demonstrated some relationship ($r > 0.3$) with the dependent variable and elimination of variables known to exhibit singularity (Pallant 2010). Relationships were determined using Pearson’s $r$ or Spearman’s $\rho$, dependent on normal distribution of data as assessed by a Shapiro-Wilks test. When these variables were identified, collinearity was identified during forced entry regression using tolerance and the variance inflation factor. Variables were considered collinear when tolerance was less than 0.10 and variance inflation factor was greater than 10 (Field 2005). When $K_{PF}$ was the dependent variable, suitable independent variables were CT, RPFD, RFD$_{0.36}$, RSI, K$_{ANK}$, PA$_{CONC}$, WA$_{CONC}$ and WA$_{ECC}$. When $K_{PF(NORM)}$ was the
dependent variable, CT, RPFD$_{\text{NORM}}$, RFD$_{0-30}$$_{\text{(NORM)}}$, K$_{\text{ANK(NORM)}}$, WA$_{\text{ECC(NORM)}}$, PS$_{\text{ECC(NORM)}}$ and PA$_{\text{CONC(NORM)}}$ were independent variables.

Both concentric and eccentric phase variables were included for initial analysis. Analysis of the duration of the eccentric phase (as defined when the ankle was dorsiflexing to its maximum point) and the timing of the peak force showed that the peak force occurred in the concentric phase (mean duration of eccentric loading: 42.2 ±6.9%, timing of peak force: 48.1±10.1%, p < 0.001, moderate effect size. The concentric variables suitable for inclusion in the models, PA$_{\text{CONC}}$ and PA$_{\text{CONC(NORM)}}$ were found to occur very late in the impact (almost 80% of CT), which is unlikely to influence an event which occurred prior to 50% of CT. Similarly, WA$_{\text{CONC}}$ is a measure of work done in the entire concentric phase rather than up to the point of interest. These variables were hence excluded from analysis.

Multivariate stepwise backwards regression was initially performed, with the equation with the lowest standard error of the estimate (SEE) selected for further analysis. From these results, variables with a non-significant t value were eliminated from analysis as this indicates a potentially zero slope of the regression line. The second model presented for each condition was created from forced entry regression using these variables with bootstrapping in SPSS (1000 samples). β was determined from this set of regression equations as it was felt to be a better indicator of the true degree of influence of an input variable compared to the initial backwards regression. The assumptions of normality, linearity and homoscedasticity were checked visually using plots of expected cumulative probability against observed cumulative probability, and a scatterplot of standardised residual against standardised predicted value.

A similar protocol was followed to identify the determinants of RSI using normalised force, force development and kinetics data. RSI was the dependent variable and F$_{\text{P(NORM)}}$, RPFD$_{\text{NORM}}$, RFD$_{0-30}$, PA$_{\text{CONC(NORM)}}$, WS$_{\text{CONC(NORM)}}$ and PS$_{\text{ECC(NORM)}}$ were included as independent variables.

To investigate what FDT and ECT may be representative of, Pearson’s r or Spearman’s ρ (dependent on normality of data) were used to assess the strength of the linear relationship between these two variables and plantarflexor function during the SSC task. Correlation coefficients were considered large when the correlation coefficient was .5 to .7 and very large when between .7 and .9 (Hopkins 2006). Simple regression was used to determine if either measure could predict K$_{\text{AT}}$ during the SSC task.
7.4 Results

Most measures were found to be highly reliable for both single and average results, with results shown in Table 11. $K_{AT}$ was very unreliable, with high CoV and low ICC scores. The next lowest reliability scores were found for the timing of $F_p$, duration of eccentric loading and $K_{ANK}$, with high observed CoV. Average ICC was still reasonably high for duration of eccentric loading and $K_{ANK}$ at .866 and .871 respectively.

The assumptions of normality, linearity and homoscedasticity were met for all equations created. Tables 12 and 13 show the results of multivariate regression with $K_{PF}$ and $K_{PF(NORM)}$ the dependent variables. All regressions were significant. High levels of accounted variance and expected variance in the population were observed with SEE of between 10.2% and 11.2% of mean $K_{PF}$ and $K_{PF(NORM)}$. $RPFD$, $RFD_{0.30}$ and CT feature consistently as important determinants in each model.

The only large correlation observed for either FDT or ECT and the SSC task variables was between FDT and the rate of tendon loading (.594). Simple regression found neither variable could predict $K_{AT}$ accurately, with SEE of 243 N.mm$^{-1}$ found using FDT and SEE of 262 N.mm$^{-1}$ using ECT as the independent variables.

Initially the 27 measures of $K_{AT}$ acquired were used in an analysis to identify the determining factors of RSI. The t value of $K_{AT}$ was statistically significant in the initial regression but was then found to be of minimal importance during forced entry regression. As a result, further analysis did not include these variables and was based on the data of 57 trials. $PS_{ECC(NORM)}$, $WS_{CONC(NORM)}$ and $PA_{CONC(NORM)}$ were found to be important determinants of plantarflexor RSI (Table 14).
<table>
<thead>
<tr>
<th>Variable</th>
<th>Group mean</th>
<th>Group standard deviation</th>
<th>Mean within-participant standard deviation</th>
<th>Mean coefficient of variation (%)</th>
<th>Cronbach’s alpha</th>
<th>Intra-class correlation coefficient</th>
<th>Single</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>CT (s)</td>
<td>0.235</td>
<td>0.051</td>
<td>0.011</td>
<td>4.7</td>
<td>0.98</td>
<td>0.93</td>
<td>0.98</td>
<td>0.98</td>
</tr>
<tr>
<td>Fp (N)</td>
<td>614</td>
<td>192</td>
<td>30</td>
<td>5.2</td>
<td>0.99</td>
<td>0.97</td>
<td>0.99</td>
<td>0.99</td>
</tr>
<tr>
<td>Time of Fp (%CT)</td>
<td>48.1</td>
<td>7.6</td>
<td>6.1</td>
<td>13.2</td>
<td>0.66</td>
<td>0.36</td>
<td>0.63</td>
<td></td>
</tr>
<tr>
<td>Duration</td>
<td>42.2</td>
<td>6.2</td>
<td>3.4</td>
<td>8.2</td>
<td>0.86</td>
<td>0.68</td>
<td>0.87</td>
<td></td>
</tr>
<tr>
<td>eccentric loading</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>RPFd (N.s⁻¹)</td>
<td>6145</td>
<td>3035</td>
<td>1070</td>
<td>19.1</td>
<td>0.94</td>
<td>0.83</td>
<td>0.94</td>
<td></td>
</tr>
<tr>
<td>RFD₀-30 (N.s⁻¹)</td>
<td>11581</td>
<td>5012</td>
<td>1875</td>
<td>18.1</td>
<td>0.91</td>
<td>0.77</td>
<td>0.91</td>
<td></td>
</tr>
<tr>
<td>RFD₀-50 (N.s⁻¹)</td>
<td>8776</td>
<td>3578</td>
<td>925</td>
<td>11.3</td>
<td>0.96</td>
<td>0.89</td>
<td>0.96</td>
<td></td>
</tr>
<tr>
<td>FT (s)</td>
<td>0.474</td>
<td>0.082</td>
<td>0.018</td>
<td>3.7</td>
<td>0.97</td>
<td>0.92</td>
<td>0.97</td>
<td></td>
</tr>
<tr>
<td>RSI</td>
<td>0.58</td>
<td>0.24</td>
<td>0.05</td>
<td>8.9</td>
<td>0.97</td>
<td>0.90</td>
<td>0.97</td>
<td></td>
</tr>
<tr>
<td>KPF (N.m⁻¹)</td>
<td>9720</td>
<td>4285</td>
<td>1064</td>
<td>11.4</td>
<td>0.97</td>
<td>0.92</td>
<td>0.97</td>
<td></td>
</tr>
<tr>
<td>PS Ecc (W)</td>
<td>340</td>
<td>150</td>
<td>35</td>
<td>10.9</td>
<td>0.98</td>
<td>0.92</td>
<td>0.97</td>
<td></td>
</tr>
<tr>
<td>PS Conc (W)</td>
<td>340</td>
<td>145</td>
<td>31</td>
<td>8.8</td>
<td>0.97</td>
<td>0.92</td>
<td>0.97</td>
<td></td>
</tr>
<tr>
<td>WS Ecc (J)</td>
<td>50</td>
<td>17</td>
<td>4</td>
<td>8.1</td>
<td>0.98</td>
<td>0.93</td>
<td>0.98</td>
<td></td>
</tr>
<tr>
<td>WS Conc (J)</td>
<td>56</td>
<td>22</td>
<td>5</td>
<td>9.7</td>
<td>0.96</td>
<td>0.90</td>
<td>0.96</td>
<td></td>
</tr>
<tr>
<td>KAT (N.mm⁻¹)</td>
<td>386</td>
<td>320</td>
<td>195</td>
<td>47.7</td>
<td>0.25</td>
<td>0.15</td>
<td>0.27</td>
<td></td>
</tr>
<tr>
<td>Rate of AT loading (N.s⁻¹)</td>
<td>32381</td>
<td>12263</td>
<td>4193</td>
<td>13.5</td>
<td>0.96</td>
<td>0.92</td>
<td>0.96</td>
<td></td>
</tr>
<tr>
<td>KANK (N.m.deg⁻¹)</td>
<td>6.76</td>
<td>6.35</td>
<td>2.07</td>
<td>24.4</td>
<td>0.88</td>
<td>0.69</td>
<td>0.87</td>
<td></td>
</tr>
<tr>
<td>PA Ecc (W)</td>
<td>354</td>
<td>100</td>
<td>31</td>
<td>10.0</td>
<td>0.96</td>
<td>0.89</td>
<td>0.96</td>
<td></td>
</tr>
<tr>
<td>PA Conc (W)</td>
<td>368</td>
<td>98</td>
<td>32</td>
<td>8.7</td>
<td>0.94</td>
<td>0.84</td>
<td>0.94</td>
<td></td>
</tr>
<tr>
<td>WA Ecc (J)</td>
<td>53</td>
<td>22</td>
<td>5</td>
<td>12.3</td>
<td>0.97</td>
<td>0.92</td>
<td>0.97</td>
<td></td>
</tr>
<tr>
<td>WA Conc (J)</td>
<td>72</td>
<td>28</td>
<td>8</td>
<td>9.8</td>
<td>0.91</td>
<td>0.78</td>
<td>0.91</td>
<td></td>
</tr>
<tr>
<td>FDT (s)</td>
<td>93</td>
<td>27</td>
<td>20</td>
<td>23.0</td>
<td>0.92</td>
<td>0.58</td>
<td>0.92</td>
<td></td>
</tr>
<tr>
<td>ECT (s)</td>
<td>56</td>
<td>10</td>
<td>5</td>
<td>9.0</td>
<td>0.96</td>
<td>0.75</td>
<td>0.96</td>
<td></td>
</tr>
</tbody>
</table>
Table 12. Predictors of absolute plantarflexor stiffness (** indicates p < 0.01)

<table>
<thead>
<tr>
<th></th>
<th>Unstandardised coefficient</th>
<th>Standard error</th>
<th>Standardised coefficients β</th>
<th>t</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>All suitable independent variables:</strong> F = 129**, R = .969, R² = .939 adjusted R² = .932, SEE = 1129 N.m⁻¹</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Constant</td>
<td>13749.543</td>
<td>1593.569</td>
<td>8.628**</td>
<td></td>
</tr>
<tr>
<td>CT</td>
<td>-41602.875</td>
<td>5883.148</td>
<td>-0.494</td>
<td>-7.072**</td>
</tr>
<tr>
<td>RPFD</td>
<td>0.502</td>
<td>0.117</td>
<td>0.368</td>
<td>4.272**</td>
</tr>
<tr>
<td>RFDₐ₀.₃₀</td>
<td>201.587</td>
<td>59.845</td>
<td>0.192</td>
<td>3.368**</td>
</tr>
<tr>
<td>RSI</td>
<td>-1651.772</td>
<td>848.495</td>
<td>-0.092</td>
<td>-1.947</td>
</tr>
<tr>
<td>K_ANK</td>
<td>99.164</td>
<td>27.254</td>
<td>0.161</td>
<td>3.639**</td>
</tr>
<tr>
<td>WA Ecc</td>
<td>19.436</td>
<td>9.688</td>
<td>0.100</td>
<td>2.006</td>
</tr>
</tbody>
</table>

Using only significant t values: F = 208**, R = .970, R² = .941, adjusted R² = .937, SEE = 1089 N.m⁻¹

<table>
<thead>
<tr>
<th></th>
<th>Unstandardised coefficient</th>
<th>Standard error</th>
<th>Standardised coefficients β</th>
<th>t</th>
</tr>
</thead>
<tbody>
<tr>
<td>Constant</td>
<td>13172.575</td>
<td>1462.094</td>
<td>9.009**</td>
<td></td>
</tr>
<tr>
<td>CT</td>
<td>-37035.631</td>
<td>4530.132</td>
<td>-0.440</td>
<td>-8.175**</td>
</tr>
<tr>
<td>RPFD</td>
<td>0.321</td>
<td>0.106</td>
<td>0.236</td>
<td>3.044**</td>
</tr>
<tr>
<td>RFDₐ₀.₃₀</td>
<td>0.237</td>
<td>0.053</td>
<td>0.295</td>
<td>4.436**</td>
</tr>
<tr>
<td>K_ANK</td>
<td>78.564</td>
<td>26.611</td>
<td>0.128</td>
<td>2.952**</td>
</tr>
</tbody>
</table>
Table 13. Predictors of normalised plantarflexor stiffness (** indicates p < 0.01)

<table>
<thead>
<tr>
<th>Unstandardised coefficient</th>
<th>Standardised coefficient β</th>
<th>t</th>
</tr>
</thead>
<tbody>
<tr>
<td>B</td>
<td>Standard error</td>
<td></td>
</tr>
</tbody>
</table>

All suitable independent variables: F = 185**, R = .974, R^2 = .948, adjusted R^2 = .943, SEE = 0.803 N.cm^{-3}m^{-1}

<table>
<thead>
<tr>
<th>Variable</th>
<th>Unstandardised coefficient</th>
<th>Standardised coefficient β</th>
<th>t</th>
</tr>
</thead>
<tbody>
<tr>
<td>Constant</td>
<td>9.118</td>
<td>1.031</td>
<td>8.843**</td>
</tr>
<tr>
<td>CT</td>
<td>-22.624</td>
<td>-0.347</td>
<td>-6.893**</td>
</tr>
<tr>
<td>RPFD_NORM</td>
<td>0.641</td>
<td>0.473</td>
<td>5.905**</td>
</tr>
<tr>
<td>RFD_0-30_NORM</td>
<td>0.351</td>
<td>0.431</td>
<td>6.958**</td>
</tr>
<tr>
<td>PS_ECC_NORM</td>
<td>-9.442</td>
<td>-0.308</td>
<td>-5.653**</td>
</tr>
<tr>
<td>K_ANK_NORM</td>
<td>38.895</td>
<td>0.063</td>
<td>1.667</td>
</tr>
</tbody>
</table>

Using only significant t values: F = 223**, R = .972, R^2 = .945, adjusted R^2 = .941, SEE = 0.816 N.cm^{-3}m^{-1}

<table>
<thead>
<tr>
<th>Variable</th>
<th>Unstandardised coefficient</th>
<th>Standardised coefficient β</th>
<th>t</th>
</tr>
</thead>
<tbody>
<tr>
<td>Constant</td>
<td>9.194</td>
<td>1.048</td>
<td>8.777**</td>
</tr>
<tr>
<td>CT</td>
<td>-22.964</td>
<td>-0.352</td>
<td>-6.893**</td>
</tr>
<tr>
<td>RPFD_NORM</td>
<td>0.661</td>
<td>0.488</td>
<td>6.027**</td>
</tr>
<tr>
<td>RFD_0-30_NORM</td>
<td>0.368</td>
<td>0.452</td>
<td>7.333**</td>
</tr>
<tr>
<td>PS_ECC_NORM</td>
<td>-9.630</td>
<td>-0.314</td>
<td>-5.682**</td>
</tr>
</tbody>
</table>
Table 14. Predictors of reactive strength index (normalised input values; * indicates $p < 0.05$, ** indicates $p < 0.01$)

<table>
<thead>
<tr>
<th>Unstandardised coefficient $B$</th>
<th>Standard error</th>
<th>Standardised coefficients $\beta$</th>
<th>t</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>All suitable independent variables:</strong> $F = 24^{**}, R = .840, R^2 = .705$ adjusted $R^2 = 0.676$, SEE = 0.14</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Constant</td>
<td>0.188</td>
<td>0.091</td>
<td>2.064*</td>
</tr>
<tr>
<td>$FP_{NORM}$</td>
<td>-0.337</td>
<td>0.330</td>
<td>-1.022</td>
</tr>
<tr>
<td>$RFD_{0-30(NORM)}$</td>
<td>-0.008</td>
<td>0.008</td>
<td>-1.039</td>
</tr>
<tr>
<td>$PSECC(NORM)$</td>
<td>2.023</td>
<td>0.435</td>
<td>4.653**</td>
</tr>
<tr>
<td>$WS_{CONC}(NORM)$</td>
<td>6.089</td>
<td>2.065</td>
<td>2.949*</td>
</tr>
<tr>
<td>$PA_{CONC}(NORM)$</td>
<td>-0.645</td>
<td>0.284</td>
<td>-2.271*</td>
</tr>
</tbody>
</table>

| Using only significant t values: $F = 39^{**}, R = .829, R^2 = .688$ adjusted $R^2 = 0.670$, SEE = 0.14 |
|--------------------------------|----------------|----------------------------------|----|
| Constant                      | 0.102          | 0.073                            | 1.394 |
| $PSECC(NORM)$                 | 1.355          | 0.198                            | 6.840** |
| $WS_{CONC}(NORM)$             | 6.827          | 1.776                            | 3.843** |
| $PA_{CONC}(NORM)$             | -0.683         | 0.279                            | -2.446* |

7.5 Discussion

Similar to the results of chapter 3, use of the sledge apparatus resulted in high reliability for most measures. Of note, while some variables reported similar average ICC (e.g. $WA_{CONC}$ and FDT, duration of eccentric loading and $K_{ANK}$), a large difference in standardised variability was observed. This highlights one of the limitations of use of any one individual reliability score as a sole determinant of measurement reliability and supports why both descriptive data and a range of measures of reliability and consistency of task performance are necessary to understand data collected. Of note, the rate of force development variables tended to have slightly higher standardised variability than outcome-related variables such.
as $F_p$, and the timing of the peak force was quite variable. This may be due to slight alterations in muscle activation strategy used to strike the plate. As force output was maintained at a similar level, this provides further evidence for the flexibility of the limb to adapt to maintain performance outcomes despite variations in imposed conditions. While mean $F_p$ is higher than previously reported in chapters 3 and 4, RPFD, RSI and FT are all lower. CT was of longer duration than in earlier chapters. These differences are almost certainly due to the heavier mass (30 to 40% body mass) added to the sledge for this particular group of participants. The reasons for this increase in test load are not clear, as participants were from a similar activity profile as those in earlier chapters and were equally familiar with the task.

Previous work has assumed all events up to peak force, the point where the stiffness measure was obtained, were eccentric (Jensen and Ebben 2007). Analysis of the duration of eccentric loading and timing of peak force showed this assumption to be incorrect for most participants which led to the inclusion of the concentric variables in initial analysis. However, the variables considered acceptable for inclusion occurred quite far after the measure of interest so it was reasonable to exclude them from analyses. Stiffness was measured in the eccentric phase for only 6 of the 57 trials analysed, so it is possible that another concentric variable not obtained during this analysis may also influence stiffness. Additional variables such as power at the time of peak force may be worth considering for inclusion in future work, as they may provide greater insight into the function and behaviour of the MTU at the specific time of interest.

The variables with the strongest influence on $K_{pF}$ and $K_{pF(NORM)}$ were RPFD, $RFD_{0-30}$ and CT. Stiffness increased with a decrease in CT and increase in rates of force development. These are reasonable variables to expect to be related, as previous work has shown increased stiffness with decreased CT during upright hopping (Hobara et al. 2007). The $RFD_{0-30}$ is highly dependent on muscle activation and activation strategy (Aagard et al. 2002). Similarly, the RPFD refers to the magnitude of $F_p$ divided by the time in seconds it takes to reach this value. $F_p$ is dependent on muscle size and architectural factors, but also level of muscle activation and how the muscle is activated. As the product of mass and acceleration, if the plate is decelerated rapidly force increases and time to peak force decreases, resulting in increased RPFD. This would be likely to occur in the presence of increased muscle activation. This increase is associated with increased muscle stiffness (Ford et al. 1981) so this would explain increased $K_{pF}$. 

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$K_{\text{ANK}}$ appears as an important predictor of $K_{pf}$ in the absolute $K_{pf}$ and as a non-significant predictor in the normalised $K_{pf}$ model. This is reasonable due to its reported importance in upright plyometric hopping tasks (Farley and Morgenroth 1999; Hobara et al. 2011). The mean value observed for $K_{\text{ANK}}$ is similar to values observed by other authors in hopping and running tasks (Kuitunen et al. 2011; Charalambous et al. 2012; Joseph et al. 2013). Joint stiffness refers to the ratio of peak ankle moment and change in joint angle from initial contact to peak moment. The potential error in measurement of joint moment which subsequently affects the $K_{\text{ANK}}$ and $K_{\text{AT}}$ measures (and may effect measurement reliability) must not be discounted. The marker-based method of determining centre of pressure has been shown to perform well in prediction of centre of pressure during hopping on the ground even when used to calculate ankle moment. It is possible that the method may not have performed as well on the sledge. The task involved repetitive striking of the plate, with the centre of pressure shown using high-speed footage to move from the ball of the foot towards the heel. Even though all participants were tested at a loading which was 70% of their 11 repetition maximum, it appears that when striking the plate, some participants allowed the centre of pressure to move much closer to the calcaneus than others. As moment was calculated as the product of force and perpendicular distance from the point of application of the force to the fulcrum (malleolus), this resulted in a number of unusual double peaked moment curves for some individuals when the product of these values dipped in the middle of the impact where the centre of pressure was close to the malleolus. The peak moment was then taken from the first peak. This is in contrast to the moments observed during running or hopping, where a smooth moment curve is observed. This observation suggests decreased ecological validity of the task in participants where this occurred. However, the potential cumulative error in joint moment calculation due to use of a surrogate measure of centre of pressure must also be considered and future work should consider both these factors during use of the adapted force sledge for calculation of ankle kinetics.

The omission of $K_{\text{AT}}$ must also be noted as the literature suggests this is a variable to consider in the modulation of $K_{\text{ANK}}$. A smaller scale analysis using the 27 suitable data points found $K_{\text{AT}}$ was included in the initial model creation, but the t value was not statistically significant and the standardised coefficient low (< 0.1). In addition, the number of data points used and reliability of the measure was also found to be very low. These
initial results suggest that $K_{AT}$ is less important than previously thought but future work with full, reliable data sets is required to fully determine the influence of this variable.

The small number of participants and cases in this analysis must be considered when considering these findings. The 57 data points used for each analysis is below the recommended $50 + 8k$ (where $k$ is the number of data points) or $104 + k$ (Field 2005). While this exploratory study provides an indication of the important factors influencing $K_{pf}$, an increased sample size is necessary for generalisation to the population.

The results show neither voluntary FDT nor ECT of soleus is related to or can accurately predict Achilles tendon stiffness. FDT is the same measure as that commonly referred to in the literature as electromechanical delay (Kubo et al. 2005b; Waugh et al. 2013). This result is surprising, as EMD has commonly been thought to reflect musculotendinous structures and indicative of the length of time taken to stretch the tendinous structures (Cavanagh and Komi 1979). Previous work has shown gastrocnemius EMD to explain 68% of variance of Achilles tendon stiffness when simultaneous measures of EMD and stiffness were obtained (Waugh et al. 2013). As measures were obtained at different times (during a heel-lift task and during a fSSC task), this may be a potential reason why no relationships were observed. It is possible that FDT or ECT in this condition is reflective of $K_{AT}$ but only in the heel-lift task. Achilles tendon stiffness has been shown to be dependent on the rate of loading on the tendon (Paxton and Baar 2007). The rate of loading in the two test conditions is very different, with a mean rate of loading of $32,000 \text{ N.s}^{-1}$ observed during the SSC task. While not calculated, the rate of loading during the heel-lift task would be expected to be much lower. Rate of loading on the Achilles tendon has been shown to be under $10,000 \text{ N.s}^{-1}$ during a walking task so would certainly be much lower during a seated heel-lift. It must be remembered that tendon behaviour during dynamic, SSC tasks is of primary interest in rehabilitation. Monitoring of tendon mechanics is important but if a surrogate measure does not reflect the stiffness during the task of interest then it may be necessary to reconsider if that ‘surrogate measure’ is suitable at all. The implication of this for the practitioner is that it may be extremely difficult to easily obtain a measure of tendon stiffness during a SSC task.

RSI was not related to tendon stiffness or tendon rate of loading which has implications for what is reported in the literature and suggestions that RSI is reflective of musculoskeletal stiffness. RSI reflects the explosive nature of a jump performance, and is
the ratio of jump height to preceding contact time. The factors considered important
predictors of RSI (\(PS_{ECC(NORM)}, WS_{CONC(NORM)}\) and \(PA_{CONC(NORM)}\)) indicate that RSI is a
measure of the amount of energy absorbed and generated during the impact. RSI increased
when each of these variables increased so the measure reflects the level of loading on the
musculoskeletal system. These variables accounted for a reasonable proportion of variance
(68.8%). While these factors may be related to the underlying mechanical behaviour of the
MTU, this does not mean imply causation. This is particularly relevant for the strength and
conditioning professional, for whom caution is advised in the use and misuse of this
measure. The RSI indicates the explosiveness of an athlete, but should not be used as a
surrogate measure of underlying MTU mechanical behaviour or rate of MTU loading. The
preliminary nature of this analysis must be noted as these results have been obtained during
an isolated muscle task on a small number of participants. Greater insight would be
obtained through repetition of this analysis during a drop jump protocol using similar
variables for analysis. As several variables were removed from analysis due to collinearity,
future work should consider the potential importance of other measures of muscle
function (e.g. \(K_{AT}\), muscle activations, knee joint kinetics) during this task.

7.6 Summary

The preceding chapters have presented a series of studies which show that an adapted
force sledge apparatus can be used to reliably assess measures of force, force development
and stretch-shortening cycle function of the plantarflexors during a stretch-shortening cycle
task. Differences in these measures have been identified due to limb preference and gender,
but to gain further insight into the reasons for these observed differences requires use of
ultrasound and inverse dynamics. The structure of the sledge does not allow it to take the
weight of a force plate on the rails so in chapter six, a motion analysis method of
determining the centre of pressure was described. This chapter incorporates these findings
to use the sledge with motion analysis and ultrasound to assess the stiffness response of the
plantarflexors during a fast stretch-shortening cycle task. The rate at which peak force
develops at, rate of force development in the first 30 ms and contact time were found to be
important in determining plantarflexor stiffness during a fast SSC task which isolates the
plantarflexors. No relationship was found between electromechanical delay variables and
Achilles tendon stiffness. Energy absorbed and generated were important determinants of
RSI.
Chapter 8: Thesis conclusions, limitations and future directions
8.1 Key findings and implications of the thesis

The aims of this thesis were to develop a method of measuring plantarflexor muscle-tendon function in a dynamic yet controlled stretch-shortening cycle test environment, and to use this method to assess plantarflexor muscle-tendon unit function in healthy young adults. The key findings and implications of this thesis were:

1. An adapted force sledge could be used to reliably assess the force, force development and reactive strength index of the plantarflexors during a fast stretch-shortening cycle task. Very high levels of reliability (single ICC >0.85, average ICC >0.95) were observed for the measures of interest (peak force, rate of peak force development, contact and flight times, duration of eccentric loading, reactive strength index). Reliability of measures was similar in the non-preferred limb, but the duration of eccentric loading was less reliable probably due to poorer motor control (CoV >9.2%, single ICC <0.74, average ICC <0.89). Similar ankle kinematics to those observed in hopping and running were observed and the rate of loading on the plantarflexors was similar to that observed in the Achilles tendon during a walking task. This suggests the adapted force sledge is a suitable method of assessing plantarflexor function in a dynamic task.

2. Large asymmetries in muscle function of up to 44.6% are present even in healthy adults with no history of injury or development of injury in the following 12 months. Significant differences of moderate effect size were observed in measures of peak force, rate of peak force development, rate of force development in the first 50 ms, flight time and reactive strength index, with no difference in the duration of contact time or magnitude of eccentric loading on the muscle. A small effect size was observed in the rate of force development in the first 30 ms. These results question the 15% upper limit of asymmetry often recommended in healthy limbs and has implications for understanding lower limb coordination strategies to compensate for altered muscle function at other joints.

3. Differences in plantarflexor force, force development and mechanical behaviour exist between males and females, with some differences removed with normalisation to plantarflexor muscle volume. The difference between genders for all variables except for reactive strength index was statistically significant. Statistically significant differences with moderate effect sizes in plantarflexor peak force, peak eccentric power and eccentric work were observed, even when data
were normalised to plantarflexor muscle volume. While not statistically significant, differences in rate of peak force development and rate of force development in the first 50 ms were of small tending towards moderate, and moderate effect size. This indicates the female plantarflexor muscle-tendon unit has a reduced capacity to absorb load, which has implications for the prescription of training exercises by the strength and conditioning professional. It may also be another potential risk factor for ACL injury in the female athlete.

4. A motion analysis marker-based method can be used with reasonable accuracy to determine the centre of pressure during a two-legged hopping task at frequencies greater than 1.5 Hz. This helps to improve the measurement of CoP in dynamic tasks without the need for use of a force plate which is fixed in place.

5. Rate of peak force development, rate of force development in the first 30 ms and contact time were found to be the main determinants of plantarflexor stiffness during a fast stretch-shortening cycle task. This has implications for our understanding of the mechanical factors and measures of muscle function modulating stiffness during this task.

6. Soleus force development time and elastic charge time, and plantarflexor RSI were not found to be related to the mechanical properties of the plantarflexor muscle-tendon unit. This highlights the difficulty in the simple assessment of a complex mechanical property.

8.2 Limitations of the thesis

- The low-load nature of this test protocol must be noted. Forces generated were around one third that could be expected during an upright hopping task, and this will have an influence on the rates of force development observed. While the results are valid for this test condition, it is unknown if similar patterns (e.g. of limb function asymmetry) are observed during upright hopping or other full weight-bearing exercises.

- While participant numbers were approximately the average expected in human studies of this nature, statistical power analysis showed that for some measures with high between-participant variability (e.g. stiffness in chapter 5), over 200 participants would be required to achieve statistical significance which is not
realistically feasible. Use of effect sizes throughout the thesis gives a better indication of between-group differences for smaller groups.

- The participants of interest were all healthy, young adults who were at least recreationally active. Results of this thesis cannot be extrapolated to different populations, such as sedentary, elderly or those with clinical problems. While the thesis has provided a foundation for future study with normative data for healthy adults, it is unknown if this method could work as well in other populations e.g. paediatrics or clinical populations, who may struggle with the motor control elements of the task.

- The lack of a force plate on the sledge is problematic. While chapter 6 presented a marker based method of determining the centre of pressure using an alternative method to the FP and determined the level of error in ankle kinetics measures, it is impossible to run a similar comparison using the sledge plate as there is nothing to compare it to. The reduced sledge reaction forces, observation of the strategy used to strike the plate in relation to ankle dorsiflexion and observation of high speed footage suggests that values observed are realistic, but there is no way of truly knowing.

- A limitation to chapter 6 comparing the accuracy of the marker method to the force plate is in fact the accuracy of the force plate. The inaccuracies of force plates for measurement of centre of pressure in a dynamic condition have been reported previously. While the accuracy of the plate used was determined at construction in the US, it is extremely difficult to determine centre of pressure accuracy during a dynamic task such as hopping. Large errors were reported in the early 1990's for dynamic centre of pressure measurement and it is unlikely that force plate design and structure has improved significantly in the intervening time.

- Throughout this thesis, no electromyography data was collected during the sledge task which prevents analysis of muscle activation strategies. This was primarily due to technical difficulties with the synchronisation of systems but it is also extremely difficult to place the EMG electrodes on the muscles of interest when the ultrasound probe is in place and the area of interest is covered with adhesive bandage and ultrasound gel.

- Scaling of data to account for differences in body size is necessary to determine if meaningful differences truly exist (Batterham et al. 1997). While scaling to body
mass is often used in the literature, caution has been advised in relation to the use of body mass as the normalisation variable in isolated muscle tasks, as there is no definitive relationship between increased body mass and increased specific muscle mass (Nevill et al. 2004). The regression method of calculating muscle volume performs well in its prediction of muscle volume but may be a potential source of error in results.

8.3 Directions of future research

This thesis has developed a controlled method of assessing the isolated plantarflexors during a stretch-shortening cycle task. While this thesis has addressed a series of related research questions, it has also raised important issues which require further investigation.

- Future work should consider the inter-day reliability of force sledges. While this thesis has established the intra-session reliability of the adapted force sledge, future work should determine if a learning effect exists when using this method.
- The relationship between measures of isolated muscle function and muscle function with full limb loading is necessary to determine how well observed measures relate to performance during tasks such as hopping or running.
- The results of chapter 4 are of particular interest, as they suggest the lower limb has a large adaptive capacity to maintain symmetrical movement during a dynamic task. The relationship between observed asymmetries in this task and with full loading is worthy of further investigation. How well these asymmetries relate to observed differences in walking and running is also of importance. Longitudinal prospective studies are required to determine the upper limit of asymmetry beyond which the risk of injury increases. This potentially varies with the nature of the sport, e.g. sprint or endurance events. The level of asymmetry at the other joints of the lower limb is also important as this may indicate the level of compensation required for symmetrical movement. Further study of lower limb coordination with consideration to muscle activation and function is necessary to fully understand this process.
- Why males perform better in this SSC task than females is still not fully understood. Body mass differences can explain certain differences, but not all. The ability to effectively utilise the SSC appears to vary with gender and with muscle-tendon unit location, so the factors that underpin effective SSC function still
warrant investigation. The increased risk of knee injuries in females is also an important area of research, and researchers are now investigating the role of the plantarflexors in knee stability and injury prevention (Fleming et al. 2001; Sherbondy et al. 2003; Shimokochi et al. 2009; Mokhtarzadeh et al. 2013). While a lower rate of force development was observed in healthy females compared to males, it may be of interest to study this in injured individuals.

- Future work needs to further investigate the factors related to the measurement of the centre of pressure during dynamic tasks as these are a vital input in both analysis of CoP position and position change, and inverse dynamics. The recent development of a force plate with static accuracy of 0.2 mm could enhance measurement and understanding of joint kinetics and stiffness and aid in development of alternative methods of measuring centre of pressure.

- The adapted force sledge performed well for analysis of force, force development and mechanical behaviour of the plantarflexors but the lack of a force plate to validate the centre of pressure position made quantification of joint kinetics difficult. The low loading also made assessment of tendon mechanical behaviour difficult. Further work in measurement of joint kinetics is necessary to develop this area of research. The results of the final study suggest that the sledge may not be the best method to determine tendon stiffness during a SSC task but the preliminary and exploratory nature of this study must be considered. Improved force measurement techniques and adjustments to the ultrasound could enhance this measurement technique.

- Throughout this thesis, most values were determined through discrete point analysis but the use of continuous methods which analyse the patterns of the entire curve, such as principal components analysis or functional data analysis, could enhance our knowledge of the processes governing performance in these tasks. The depth of knowledge required to successfully use these analysis techniques was outside the scope of this thesis but is certainly an area which warrants further investigation.

- It is still unknown if a surrogate measure of tendon stiffness can be easily obtained, or what the force development time and elastic charge time represent. While the method to do so at the time of testing was not ready for use, use of ultrasound and
inverse dynamics during that particular protocol may provide further insight into what is really happening at tendon level.

- Similarly, the results of chapter 7 suggest a lack of clarity in what the reactive strength index actually represents. This has implications for both the strength and conditioning professional, and those involved in strength and conditioning research. While this thesis has shown that it is primarily determined by energy absorption and generation during the dynamic task, further work is required to investigate this variable during jumping tasks.
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