LOWER LIMB MECHANICAL CHARACTERISTICS DURING DYNAMIC TASKS AND PATELLAR TENDON MECHANICAL PROPERTIES IN ELITE ATHLETES

Submitted by

Amitabh Gupta

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School of Physiotherapy

Faculty of Health Sciences

La Trobe University

Victoria 3086

Australia

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Abbreviations

ADL	activities of daily living
AL	alternate-leg hopping
BF	biceps femoris
CI	confidence interval
COM	centre of mass
COP	centre of pressure
CSA	cross-sectional area
CV	co-efficient of variation
D	summed proximal (d1) and distal (d2) PT lengths
d1	proximal PT segment length
d2	distal PT segment length
DL	double-leg hopping
EMG	electromyography
GM	gluteus medius
GRF	ground reaction force
Hz	Hertz
IC	initial contact
k_L	vertical leg stiffness
MG	medial gastrocnemius
MRI	magnetic resonance imaging
MTU	muscle-tendon-unit
MVIC	maximal voluntary isometric contraction
PT	patellar tendon
QF	quadriceps femoris
SD	standard deviation
sEMG	surface electromyography
SEC	series elastic component
SSC	stretch-shorten-cycle
TA	tibialis anterior
T_f	time in flight phase
T_c	time in contact phase
TL	total patellar tendon length
TO	toe-off
vGRF	vertical ground reaction force
VAS	visual analogue scale
VISA	Victorian Institute of Sports Assessment
VL	vastus lateralis

z_c vertical height displacement in braking phase

z_f vertical height displacement in flight phase

Symbols

*	$p \leq 0.05$
**	$p \leq 0.01$

Glossary of terms

<u>Elastic Modulus</u>
In this thesis, elastic modulus represents the gradient or slope of a part of the complete stress-strain curve of the patellar tendon. In the current thesis the two points that form a straight line (slope) represent a 10% increment or range of effort of the maximal voluntary isometric contraction effort of the knee extensors.

<u>Hopping</u>
Hopping refers to the cyclical, rhythmical and repeated performance of hops. A hop is composed of a flight phase and the subsequent contact phase. The primary direction of displacement of the centre of mass during hopping is along the vertical axis.

<u>Jumping</u>
In this thesis, jumping refers to a single effort to displace the centre of mass in the vertical and or horizontal direction.

<u>Lower Limb Mechanical Behaviour</u>
This term encompasses the mechanics of the human body modelled as a spring-mass, lower limb mechanical behaviour, joint mechanical behaviour and muscle activation characteristics.

<u>Stiffness</u>
(a) *Vertical Leg Stiffness* represents the quotient of the change in vertical ground reaction force to vertical height displacement of the centre of mass during the braking phase.
(b) *Joint Stiffness* represents the quotient of joint moment and joint excursion during the braking phase.
(c) *Incremental Stiffness* represents the gradient or slope of a part of the complete force-length curve of the patellar tendon. In the current thesis the two points that form a straight line (slope) represent a 10% increment or range of effort of the maximal voluntary isometric contraction effort of the knee extensors.

<u>Stretch –Shorten-cycle</u>
Action of the muscle-tendon-unit that includes an eccentric then concentric muscle action.

Summary

Elite level jumping athletes have a high prevalence of patellar tendinopathy. A number of factors are associated with the development of this often chronic and debilitating condition. Primarily, the performance of repeated loading at the knee joint is implicated in the development of overuse injury at the knee. The ability of the lower limb to withstand the kinetic energy generated during repeated stretch-shorten-cycle (SSC) loading tasks requires a co-ordinated and adaptable response between the knee and ankle joints. Specifically at the knee, the patellar tendon (PT) may adapt its mechanical properties to accommodate any changes in load. Therefore, the aims of this thesis were; to evaluate the role of the knee and ankle joints in elite jumping athletes during a repeated loading task and to evaluate the PT mechanical properties in groups of jumping and non-jumping athletes.

Repeated lower limb loading during hopping under different conditions was used to assess the differences in mechanical and muscle activation characteristics of the lower limb. This study determined that the ankle joint was the primary site at which mechanical behaviour changed when hopping conditions changed. However, when ankle function was constrained by hopping on a decline surface there was a concomitant increase in mechanical work done at the knee joint that allowed for maintenance of hopping and lower limb function. The findings of this study implicated an increase in the capacity of the muscles acting about the knee joint when ankle function was reduced.

Validity and reliability of a method to measure PT elongation, as a sum of proximal and distal PT segment lengths during a maximal voluntary isometric contraction of the knee extensors was established. This was necessary as the available ultrasound technology did not allow complete visualisation of the PT in taller individuals. Subsequently, cross-sectional and prospective studies measuring PT mechanical properties *in vivo* were performed for groups of elite level athletes.

The cross-sectional study determined that PT mechanical properties including stiffness and elastic modulus were similar for groups of athletes across different sports (volleyball, basketball and swimming) and genders (basketball). A significant difference in PT stiffness between men’s volleyball players based on their preferred landing leg, suggests that there

may be an association between PT mechanical behaviour and dominance-related landing strategy. A follow-up of 13 male jumping athletes one year later demonstrated that there was an increase in PT mechanical work done and decrease in elastic modulus values. The findings of these two studies suggest that the PT changed its tolerance to tendon deformation under load that was associated with the dominance related landing strategy in men’s volleyball players and with chronic exposure to loading in jumping athletes.

The findings of this thesis demonstrate the task-specific nature of adaptation during dynamic and static testing procedures. Specifically, the decline surface during hopping resulted in a reduction in ankle joint function with a concomitant increase at the knee joint during hopping on a decline surface. A change in PT mechanical properties was specific to the loading strategy that the PT was exposed to rather than the health of the PT or type of sports participation of the athlete. These findings demonstrate the ability of the musculoskeletal system to adapt to changes in lower limb function.

Statement of Authorship

Except where reference is made in the text of the thesis, this thesis contains no material published elsewhere or extracted in whole or in part from a thesis submitted for the award of any other degree or diploma.

No other person's work has been used without due acknowledgement in the main text of the thesis.

This thesis has not been submitted for the award of any degree or diploma in any other tertiary institution.

The research procedures reported in the thesis were approved by the La Trobe University, Faculty of Health Sciences, Human Ethics Committee and the Australian Institute of Sport Human Research Ethics Committee.

Signed:…………………………………………………..   Date:…………………………..

Note: This thesis has been prepared to conform to the La Trobe University (Schedule C: presentation of a thesis) guidelines and is based on the style recommended in the publication manual of the American Psychology Association. Spelling conforms to Australian English.

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List of Conference Presentations

Gupta A, Purdam C, Cook J, Allison GT, ‘Comparison of patellar tendon mechanical properties in jumping and non-jumping elite athletes’ Australian Physiotherapy Association National Conference, oral presentation, October 2009, Sydney Conference Centre.

Gupta A, Purdam C, Cook J, Allison GT, ‘A comparison of two methods for calculation of vertical leg stiffness during hopping’ Australian Physiotherapy Association National Conference, oral presentation, October 2009, Sydney Conference Centre.

Gupta A, Allison GT, Cook J and Purdam C, ‘Evaluation of patellar tendon mechanical properties and preferred landing leg in elite jumping athletes’ International Society of Biomechanics In Sport, International Conference, oral presentation, July 2008, Seoul National University, Seoul, South Korea.

Gupta A, Purdam C, Cook J and Allison GT, ‘Evaluation of patellar tendon mechanical properties in elite jumping athletes’ Journal of Science and Medicine in Sports 12 (Supplement 1) January 2009: S66. Sports Medicine Australia, National Conference, oral presentation, Fiji, Shangri La Resort, October 2008.

1 Chapter One: Introduction

1.1 Chapter Aims

This chapter aims to:

- outline the prevalence of patellar tendinopathy in jumping athletes.
- outline the interaction between the stretch-shorten-cycle and tendon mechanical behaviour.
- discuss the methods used to evaluate mechanical behaviour of the lower limb and patellar tendon
- state the goal and aims of this thesis.

1.2 Introduction

Evaluation of knee function in elite jumping athletes is important due to the susceptibility to overuse knee injury in these athletes (Lian, Engebretsen, & Bahr, 2005; Tiemessen, Kuijer, Hulshof, & Frings-Dresen, 2008). The pathogenesis of overuse injury is related to the exposure to repeated microtrauma rather than a particular event that may result in injury (Bahr, 2009; Fuller, Ekstrand, Junge, Andersen, Bahr, Dvorak, Hägglund, McCrory, & Meeuwisse, 2006) and implicates multiple factors. Patellar tendinopathy is the most prevalent and recalcitrant overuse knee injury in volleyball and basketball players (Cook, Khan, Harcourt, Grant, Young, & Bonar, 1997; Cook, Khan, Harcourt, Kiss, Fehrmann, Griffiths, & Wark, 1998; Ferretti, 1986; Kettunen, Kvist, Alanen, & Kujala, 2002; Lian, et al., 2005; Tiemessen, et al., 2008; Witvrouw, Bellemans, Lysens, Danneels, & Cambier, 2001).

The patellar tendon (PT) attaches to the inferior pole of the patella and the tibial tuberosity (Kramer, White, & Recht, 2009; A. H. Sonin, 1994). It transmits the tensile force developed by the quadriceps femoris (QF) to the tibia and generates a knee extensor moment. Patellar tendon tissue is subject to high loading as a consequence of the demand placed on the knee extensors during sport, especially in jumping athletes (Tiemessen, et al., 2008). Patellar tendinopathy is most common at the posterior aspect of the proximal attachment to the inferior pole of the patella (Cook, et al., 1998; Johnson, Wakeley, & Watt, 1996). Findings of differential mechanics between anterior and posterior PT bundles vary, with report of lower strain posteriorly (Almekinders, Vellema, & Weinhold, 2002; P. Hansen, Haraldsson, Aagaard, Kovanen, Avery, Qvortrup, Larsen, Krogsgaard, Kjaer, & Magnusson, 2010), greater strain posteriorly (Basso, Amis, Race, & Johnson, 2002; P. Hansen, et al., 2010), greater forces posteriorly (Dillon, Erasmus, Müller, Scheffer, & de Villiers, 2008), lower stress posteriorly (P. Hansen, et al., 2010; Haraldsson, Aagaard, Krogsgaard, Alkjaer, Kjaer, & Magnusson, 2005) and lower elastic modulus posteriorly (P. Hansen, et al., 2010; Haraldsson, et al., 2005). Furthermore, there is evidence demonstrating greater tensile strain at the site of injury (Lavagnino, Arnoczky, Elvin, & Dodds, 2008) and at the bone-tendon interface compared to the mid-substance of the human PT (Stouffer, Butler, & Hosny, 1985). Although findings conflict they implicate differential load between anterior and posterior tendon bundles that may be a risk factor for the pathogenesis of patellar tendinopathy that is most commonly located at the posterior aspect of the attachment to the patella (el-Khoury, Wira, Berbaum, Pope, & Monou, 1992; Gemignani, Busoni, Toneroni, & Scaglione, 2008; Hoksrud, Öhberg, Alfredson, & Bahr, 2008; McLoughlin, Raber, Vellett, Wiley, & Bray, 1995; Peers & Lysens, 2005; Pfirrmann, Jost, Pirkl, Aitzetmüller, & Lajtai, 2008).

A number of further intrinsic risk factors have been found to be associated with patellar tendinopathy including: reduced hamstring and QF flexibility (Witvrouw, et al., 2001), reduced ankle dorsiflexion (Malliaris, Cook, & Kent, 2006), an increased volume of weight training (Lian, Refsnes, Engebretsen, & Bahr, 2003), greater body mass (Crossley, Thancaamootoo, Metcalf, Cook, Purdam, & Warden, 2007; Lian, et al., 2003; Malliaras, Cook, & Kent, 2007), greater body mass index (Crossley, et al., 2007; Malliaras, et al., 2007), increased waist girth and hip girth in male volleyball players (Malliaras, et al., 2007), jumping performance (Crossley, et al., 2007; Lian, et al., 2003), greater hours of sports participation and greater thigh flexibility for athletes with bilateral patellar tendinopathy (Crossley, et al., 2007), reduced thigh strength for athletes with unilateral patellar tendinopathy (Crossley, et al., 2007) and longer inferior pole of the patella (Lorbach, Diamantopoulos, Peter-Kammerer, & Paessler, 2008). It remains unknown how the loading behaviour of the PT may implicate the pathogenesis of patellar tendinopathy. This thesis aims to evaluate lower limb mechanical behaviour that may lead to an increase in load of the knee joint and how potential differences in PT loading profiles of different sporting groups may affect the PT mechanical properties.

Tendon mechanical properties have been shown to be sensitive to changes in mechanical loading (Hoyt, Wickler, Biewener, Cogger, & De La Paz, 2005; Kjaer, Langberg, Heinemeier, Bayer, Hansen, Holm, Doessing, Kongsgaard, Krogsgaard, & Magnusson, 2009; Lieber, Leonard, & Brown-Maupin, 2000; Maganaris, 2002; Maganaris & Paul, 2000; Magnusson, Narici, Maganaris, & Kjaer, 2008; Narici & Maganaris, 2006). The main factor that influences PT loading is the mechanical behaviour of the joints about which the muscle-tendon-unit (MTU) acts to generate or dissipate force. Loading the PT may change the mechanical and morphological characteristics of the PT as it is sensitive to adaptation (Doillon, Dunn, Bender, & Silver, 1985; Kongsgaard, Qyortrup, Larsen, Aagaard, Doessing, Hansen, Kjaer, & Magnusson, 2010; Lian, Scott, Engebretsen, Bahr, Duronio, & Khan, 2007; Parry, Barnes, & Craig, 1978; Shimomura, Jia, Niyibizi, & Woo, 2003). However, there is a paucity of literature on lower limb mechanical behaviour of elite level jumping athletes and on the mechanical behaviour of the PT in elite level jumping athletes. In this thesis the term lower limb mechanical behaviour encompasses the mechanics of the human body modelled as a spring-mass, lower limb mechanical behaviour, joint mechanical behaviour and muscle activation characteristics.

1.2.1 Stretch-Shorten Cycle and Tendon Mechanical Properties

Knowledge of the MTU mechanical behaviour has been advanced by technologies including motion analysis, electromyography (EMG) and ultrasound in addition to more traditional approaches of measurement including force plates and dynamometry. These approaches have provided insight into the role of MTU behaviour during dynamic tasks utilising the stretch-shorten-cycle (SSC) (Cavagna, 1977; Komi, 1984; Kyroläinen, Finni, Avela, & Komi, 2003; Roberts, 2002). Eccentric action in a SSC is usually fast, of short duration and controlled by central and peripheral neural pathways (Nicol, Avela, & Komi, 2006). It is accepted that a SSC can generate greater force than either a concentric or eccentric muscle action alone, primarily due to the storage and release of elastic energy (Bosco, Ito, Komi, Luhtanen, Rahkila, Rusko, & Vitasalo, 1982; Cavagna, 1977; Komi & Bosco, 1978). To this end elite jumping athletes provide the ideal population to observe and measure the mechanical and muscle activation parameters in utilisation of the SSC.

Lichtwark and Wilson (2005) determined that 16% of the total average mechanical work in the Achilles tendon during single-leg hopping was a result of returned energy and that over

80% of the length change in the gastrocnemius MTU is at the site of the Achilles tendon (Lichtwark & Wilson, 2005b). The ability of connective tissue to store and release energy is demonstrated at the Achilles tendon (Bobbert, 2001; Lichtwark, et al., 2005b; Lichtwark & Wilson, 2007), medial gastrocnemius (MG) aponeurosis (Chino, Oda, Kurihara, Ngayoshi, Yoshikawa, Kanehisa, Fukunaga, Fukashiro, & Kawakami, 2006; Fukashiro, Kurukawa, Hay, & Nagano, 2005; Fukunaga, Kubo, Kawakami, Fukashiro, Kanehisa, & Maganaris, 2001; Kawakami, Muraoka, Ito, Kanehisa, & Fukunaga, 2002; Kubo, Kanehisa, & Fukunaga, 2005a, 2005b; Kubo, Kanehisa, Takeshita, Kawakami, & Fukunaga, 2000c; Kurokawa, Fukunaga, & Fukashiro, 2001) and vastus lateralis (VL) aponeurosis (Bojsen-Møller, Hansen, Aagaard, Kjaer, & Magnusson, 2003; Bojsen-Møller, Magnusson, Rasmussen, Kjaer, & Aagaard, 2005; De Zee & Voight, 2001; Finni, Ikegawa, Lepola, & Komi, 2003; Ishikawa & Komi, 2004; Ishikawa, Komi, Finni, & Kuitunen, 2006; Kawakami, Kubo, Kanehisa, & Fukunaga, 2002; Kubo, et al., 2005a).

Investigation of the mechanical behaviour of the MTU during a SSC demonstrates that rapid shortening of tendon tissue is possibly due to the contractile component maintaining a ‘near-isometric’ force-length relationship during walking and jumping (Belli & Bosco, 1992; Finni, et al., 2003; Fukashiro, Hay, & Nagano, 2006; Fukunaga, Kawakami, Kubo, & Kanehisa, 2002a; Fukunaga, et al., 2001; Griffiths, 1991; Hof, Geelen, & van den Berg, 1983; Ishikawa, Finni, & Komi, 2003; Ishikawa, Komi, Grey, Lepola, & Bruggemann, 2005; Kurokawa, Fukunaga, Nagano, & Fukashiro, 2003; Magnusson, et al., 2008), This near-isometric behaviour is illustrated in the length-tension curve (Figure 1.2.1) whereby during dynamic activities there is minimal change in length of the sarcomere during the dynamic activity. This ‘near-isometric’ or ‘quasi- isometric’ behaviour of the contractile component is also demonstrated in animals across a range of contraction velocities (Biewener, Konieczynski, & Baudinette, 1998; Burkholder & Lieber, 2001; Griffiths, 1991; Hoffer, Caputi, Pose, & Griffiths, 1989; Roberts, Marsh, Weyand, & Taylor, 1997; Zuurbier & Huijing, 1992).

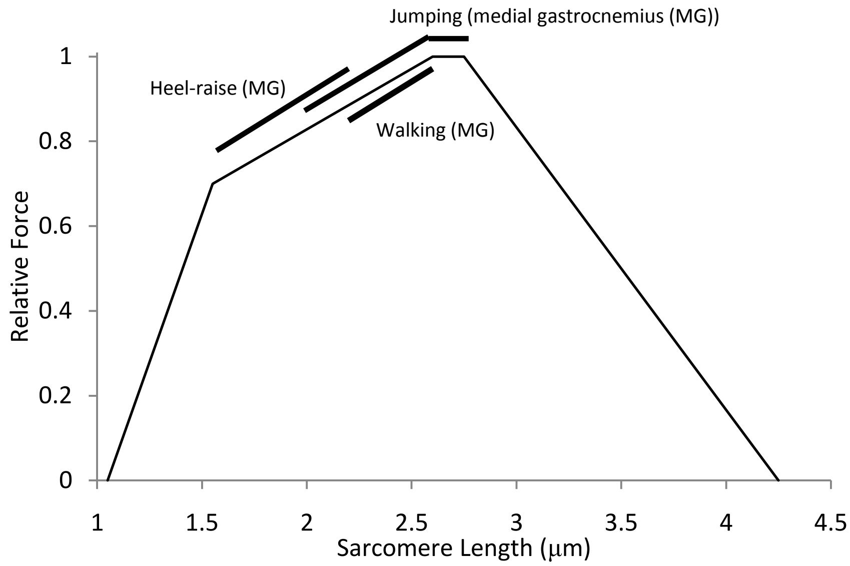


Figure 1.2.1 Force-length curve for a single sarcomere (Fukunaga, et al., 2002a). This figure illustrates that for the medial gastrocnemius muscle during heel-raise (Kubo, et al., 2000c), jumping (Kurokawa, et al., 2001) and walking (Fukunaga, et al., 2001) the sarcomere length maintains a length where the force is optimal throughout the function.

Stiffness (force/deformation) represents the elasticity of the tissue (Viidik, 1972) and hysteresis is an indication of the viscous properties (D. L. Butler, Grood, Noyes, & Zernicke, 1978). Optimal stiffness (more stored elastic energy) and hysteresis (greater re-used elastic energy) favour the performance of the SSC (Kubo, et al., 2005b). Tendon stiffness at the Achilles tendon has been related to the capacity to jump high (Bobbert, 2001) and at the knee extensors has been related to jumping and sprinting performance (Kubo, Kanehisa, Kawakami, & Fukunaga, 2000a, 2000b; Kubo, Kawakami, & Fukunaga, 1999). Tendons with reduced stiffness can more efficiently store and utilise energy during the SSC in humans (Cavagna, 1977; Kubo, et al., 2005b; Kubo, et al., 2000a, 2000b; Kubo, et al., 2000c; Kubo, et al., 1999; Walshe, Wilson, & Murphy, 1996). However, the time available for re-utilisation of stored energy is less in dynamic and relatively fast versus slow contractions, as there is a reduction in the toe-region and linearity of the force-length curve (Fukashiro, et al., 2006) (Figure 1.2.2). Thus, the performance of the activity must be relatively fast for efficient use of stored energy in tendon tissue (Fukashiro, et al., 2006; Fukashiro, et al., 2005). Furthermore, there is an optimal stiffness of the MTU or lower limb when modelled as a spring. Stiffness above or below this optimal value for a given load will affect the release of energy.

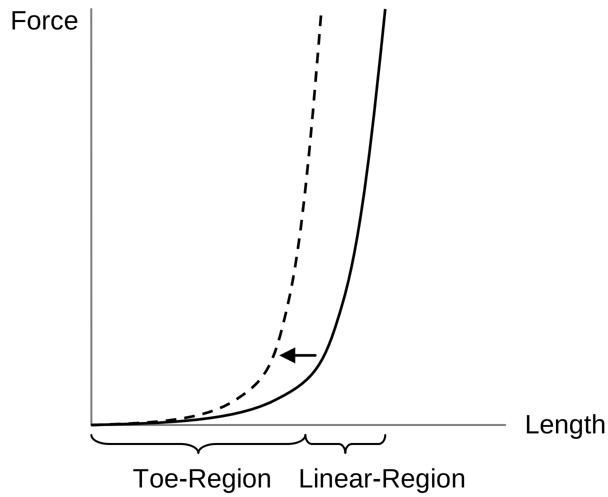


Figure 1.2.2 Schematic representation of the force-length relationship for tendon highlighting the toe and linear regions adapted from (Buckwalter, Einhorn, & Simon, 2000). There may be a reduction in the toe-region (arrow) in fast compared to slow activities.

Mechanical efficiency of the body during hopping and running was determined to be sensitive to changes in tendon mechanical properties of the patellar and Achilles tendons (M. Voight, Bøjsen-Møller, & Simonsen, 1995a). Voight et al. (1995) also demonstrated that a significant portion of the total mechanical efficiency was due to tendons, with a greater enhancement of work compared to the muscles during maximal vertical landing and repetitive skipping (M. Voight, et al., 1995a; M. Voight, Simonsen, Dyhre-Poulsen, & Klausen, 1995b). Importantly, Voight et al. reported that when values for Achilles tendon cross-sectional area (CSA) and elastic modulus were either optimised or reduced by 5% in the model for a single subject, their contribution to a change in the mechanical efficiency of the muscle-tendon complex was not as profound as when the Achilles tendon moment arm was altered by ±10% (M. Voight, et al., 1995a). Based on this finding, Voight et al. asserted that differences in moment arm may explain differences in the economy of movement between individuals (M. Voight, et al., 1995a).

Although both the Achilles and PT are susceptible to tendinopathy, the factors that may influence the mechanical behaviour of each tendon may be different. The Achilles tendon is long (ranging from 11 to 26cm) and spreads from the gastrocnemius muscle proximally (4.5 to 8.6cm in width) narrowing to a more rounded shape distally (1.2 to 2.6cm in width) before inserting at the calcaneus with a crescent shaped attachment (2.0 to 4.8cm in width) (Doral, Aklam, Bozkurt, Turhan, Atay, Dönmez, & Maffulli, 2010). This is in contrast to the PT which is relatively shorter (~5cm), more broad (2 to 3cm) and more flat (5 to 6mm in thickness) (A.H. Sonin, Fitzgerald, Bresler, Kirsch, Hoff, & Friedman, 1995). The ability to store and utilise energy is governed by the tendon compliance (inverse of stiffness) and its length. A MTU with a relatively longer series elastic component (SEC) will have greater elasticity and will be better able to store and release elastic energy (Biewener, et al., 1998; Roberts, Chen, & Taylor, 1998; Roberts, et al., 1997) than a MTU with a relatively shorter SEC such as the knee extensors (Alexander & Ker, 1990; Fukashiro, et al., 2006; Gillis & Biewener, 2001; Sawicki, Lewis, & Ferris, 2009). The potential to store energy is influenced by the strain of the tendon; Achilles tendon i*n vivo* strain has been shown to range from a mean of 4.4% (SEM) (0.3) (Arya & Kulig, 2010; Muramatsu, Tetsuro, Takeshita, Kawakami, Hirano, & Fukunaga, 2001) to 8% (1.2) (Magnusson, Hansen, Aagaard, Brønd, Dyhre-Poulsen, Bojsen-Møller, & Kjaer, 2003a) and PT strain from 4.8% (1.6) (Westh, Kongsgaard, Bøjsen-Møller, Aagaard, Hansen, Kjaer, & Magnusson, 2008) to 16.5% (1.9) (Onambélé, Burgess, & Pearson, 2007a).

Although the variability in measures may be due to differences in methods, difference in strain will directly impact on the tendon’s ability to store energy. The key feature in determining the performance of displacing a mass (i.e.; height of displacement) with respect to energy storage was identified as the ratio of length of the SEC to the contractile component (Nagano, Komura, & Fukashiro, 2004). For light objects the height of displacement was greater when the SEC was longer, and for heavier objects the height of displacement was greater when the SEC was relatively shorter (Nagano, et al., 2004). This model fits the comparison between the Achilles and patellar tendons, with a greater ratio of SEC to contractile component for the Achilles tendon and smaller ratio for the PT (Alexander & Bennet-Clark, 1977; Ker, 2002; Ker, Alexander, & Bennett, 1988). Considering that there is an association between athletes who jump higher to be at greater risk of developing patellar tendinopathy (Lian, Engebretsen, Ovrebø, & Bahr, 1996; Reeser, Verhagen, Briner, Askeland, & Bahr, 2006), greater PT load may be implicated in the pathogenesis of patellar tendinopathy.

The PT is susceptible to overload and pathology in volleyball and basketball players where it affects approximately 50% of male players (Blazina, Kerlan, Jobe, Carter, & Carlson, 1973; Ferretti, 1986; Ferretti, Puddu, Mariani, & Neri, 1984; Kujala, Kvist, & Osterman, 1986; Lian, et al., 2005; Reeser, et al., 2006). It remains unclear why some athletes experience pain and others do not with similar training loads (Reeser, et al., 2006), especially with a finding that there are often identifiable ultrasound changes without the presence of pain (Cook, Feller, Bonar, & Khan, 2004a; Cook, et al., 1998; Cook, Khan, Kiss, Coleman, & Griffiths, 2001a; Cook, Khan, Kiss, Purdam, & Griffiths, 2000b; Gisslen & Alfredson, 2005).

1.2.2 Evaluation of the Knee Joint and Patellar Tendon

Tendon mechanical behaviour changes with respect to function (Chino, et al., 2006; Fukashiro, et al., 2006; Fukunaga, Kawakami, Muraoka, & Kanehisa, 2002b; Fukunaga, et al., 2001; Kubo, et al., 2000c; Kubo, et al., 1999; Lichtwark, Bougoulias, & Wilson, 2007; Nagano, et al., 2004; M. Voight, et al., 1995a). The focus of this thesis was specifically on elite jumping athletes who are susceptible to patellar tendinopathy. Measurement of mechanical and muscle activation characteristics during dynamic hopping tasks were made to evaluate the role of the knee during the SSC. At the local tissue level, the mechanical behaviour of the PT was evaluated to identify characteristics that may be unique to athletes based on their sport or the presence of tendinopathy. Athletes participating in different sports are exposed to different loading profiles of the PT and hence may demonstrate differences in PT mechanical properties. This is analogous to differences in the morphology of the tibia between recruits that participate in activities with different exercise loading patterns (Nikander, Kannus, Rantalainen, Uusi-Rasi, Heinonen, & Sievänen, 2009; Rantalainen, Nikander, Heinonen, Suominen, & Sievänen, 2010).

Commonly available diagnostic ultrasound systems do not currently allow a non-invasive and readily available method to determine *in vivo* PT mechanics during rapid movements. Elite level athletes in this investigation could not volunteer for testing procedures that may risk their health or well-being with invasive methods such as optic fibre insertion (Dillon, et al., 2008; Elvin, Elvin, Scheffer, Arnocsky, Dillon, & Erasmus, 2009; Finni, Komi, & Lepola, 2001). Therefore, lower limb mechanical behaviour and muscle activation was determined during dynamic tasks and PT mechanical behaviour was determined using static measures. The method used to measure PT mechanics allowed for a short testing period and did not place participants at risk of experiencing pain or injury due to the testing method.

1.3 Research Goal and Aims

The purpose of this thesis was to identify lower limb and tissue based mechanical characteristics in elite jumping athletes with and without patellar tendinopathy.

Key research aims were to:

- identify the changes and evaluate the interaction between knee and ankle joints in lower limb mechanical behaviour during hopping across different conditions.
- compare mechanical properties of the patellar tendon between elite athletes from different jumping and non-jumping sports.
- evaluate the change in mechanical properties of the patellar tendon of elite jumping athletes over a season of training and competition.
- compare the inter-side differences in lower limb and patellar tendon mechanical behaviour of elite jumping athletes.

1.4 Hypotheses

This thesis tested the following hypotheses:

1. That there is no difference in joint mechanical behaviour between hopping tasks under different conditions for elite level jumping athletes.
2. That there is no difference in patellar tendon mechanical properties between elite level athlete groups participating in jumping and non-jumping sports.
3. That there is no difference in patellar tendon mechanical properties between initial and follow-up testing sessions for elite level jumping athletes.
4. That there is no inter-side difference in lower limb mechanical behaviour and muscle activation characteristics or in patellar tendon mechanical properties of elite level jumping athletes.

2 Chapter Two: Background to Hopping

2.1 Chapter Aims

This chapter aims to:

- establish background knowledge of lower limb mechanical and muscle activation characteristics during hopping.

2.2 Introduction

Hopping is a repeatable and rhythmical task that utilises the SSC in lower limb muscles and involves the storage and release of energy (Alexander, et al., 1977; Anderson & Pandy, 1993; Cavagna, 1977; Ettema, Van Soest, & Huijing, 1990; Hof, et al., 1983; Huijing, 1992; Komi, 1979, 2000). Hopping consists of flight and contact phases and a complete hopping cycle is described by a flight phase and the proceeding contact phase (Figure 2.2.1).

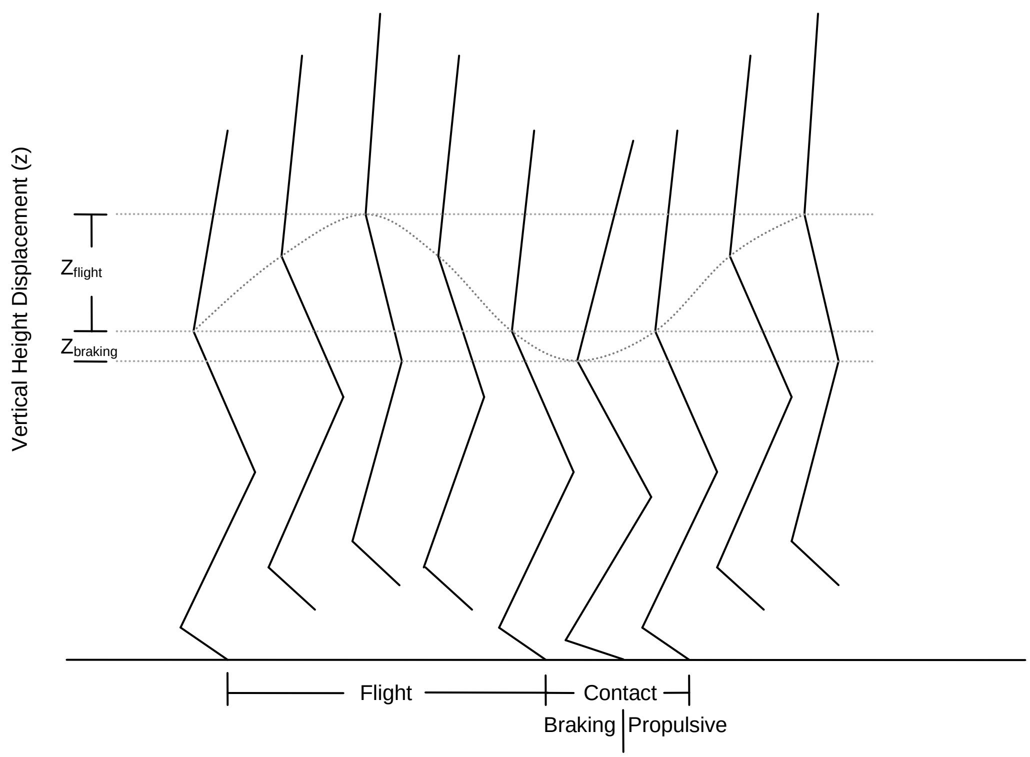


Figure 2.2.1 This schematic representation of human hopping illustrates the change in body position with respect to time as the individual aims to maintain the same position in the x and y axes (transverse plane). A hop consists of a complete flight phase and the following contact phase. The contact phase is divided into the braking and propulsive phases. Vertical height displacement represents the distance along the z axis that the centre of mass travels in flight or contact phases.

Humans have the ability to alter their hopping pace by either increasing or decreasing the hopping frequency and thus alter motor control strategy based on the task-specific nature of the skill. Hopping frequency at a self-selected pace is similar across healthy populations (~2.2Hz) (Farley, Blickham, Saito, & Taylor, 1991; Funase, Higashi, Sakakibara, Imanaka, Nishihira, & Miles, 2001; Hobara, Kanosue, & Suzuki, 2007; Melvill-Jones & Watt, 1971; Padua, Carcia, Arnold, & Granata, 2005) and has a narrow range of natural variability due to physiological constraints of force development, displacement of the centre of mass (COM), MTU stiffness and body mass (Blickhan, 1989; Farley, Blickhan, & Taylor, 1985; Heglund & Taylor, 1988).

Instructing participants on how to perform drop jump tasks has been shown to influence contact time, altering leg and ankle stiffness (Arampatzis, Schade, Walsh, & Brüggemann, 2001) as well as changing EMG activation patterns in a series of run and stop (single-leg) tasks (Cowling, Steele, & McNair, 2003). Similarly, studies have used verbal instructions to assess double-leg (DL) hopping on a flat surface including instructions to 'hop higher', to 'hop faster' or to 'hop as high as possible' in order to modulate the hopping frequency (Farley, et al., 1991; Farley & Morgenroth, 1999; Hobara, Muraoka, Omuro, Gomi, Sakamoto, Inoue, & Kanosue, 2009); matching the hopping frequency to an audible metronome has aimed to reduce the variability in hopping frequency (Farley, et al., 1991; Funase, et al., 2001; Hobara, Inoue, Muraoka, Kohei, Sakamoto, & Kanosue, 2010a; Hobara, Kimura, Kohei, Gomi, Muraoka, Iso, & Kanosue, 2008; Moritz & Farley, 2005; Padua, Arnold, Perrin, Gansneder, Carcia, & Granata, 2006); maintaining a rigid knee has aimed to reduce the influence of the knee on changes in leg stiffness (Hobara, et al., 2007; Lloyd, Oliver, Hughes, & Williams, 2009). Other factors that have been investigated to assess a change in leg mechanics during hopping are to hop to fatigue (Maton & Le Pellec, 2001) and to single-leg hop (Austin, Garrett, & Tiberio, 2002a; Austin, Tiberio, & Garrett, 2002b, 2003; Chang, Roiz, & Auyang, 2008; Lichtwark, et al., 2005b; Melvill-Jones, et al., 1971). These studies have investigated differences in kinetic, kinematic and muscle activation characteristics. Extrinsic factors such as the use of an ankle-foot orthosis (Chang, et al., 2008) or alteration of the stiffness of the hopping surface (Ferris & Farley, 1997; Moritz, et al., 2005) have also been used to evaluate kinetic, kinematic and muscle activation adjustments of the lower limb.

Kinetic and kinematic parameters represent the ‘output’ of a motor control strategy. Features that may change in the kinetics and kinematics are primarily the result of changes to the stiffness of the joints. The joints are primarily controlled by stiffness of the MTUs crossing the joint. The change in MTU stiffness is dependent on the force, velocity and time of contraction, that in hopping, is task-specific. Tensile force is produced through both the contractile component and SEC that have primarily included an investigation of the quadriceps femoris and gastrocnemius muscles during hopping (Fukashiro, et al., 2006; Fukunaga, et al., 2002a). Electromyography provides a relative measure of contractile component neural activity and allows analysis of a change in level of muscle activation.

2.2.1 Knee and Ankle Motor Control during Hopping

The interaction between the ankle and knee during hopping has been an area of recent research. Hobara et al. (2010) determined a change in knee and hip stiffness when hopping was conducted over a range of frequencies (1.5, 2.2 and 3.0Hz) with no change in ankle joint stiffness (Hobara, et al., 2010a). Hobara et al. (2010a) demonstrated that a change in kinematics rather than neural activity adjusted leg and joint (hip and knee) stiffness as an adaptation to hopping at different frequencies. The findings by Hobara et al. (2010) conflict with that of earlier studies that demonstrated ankle stiffness to be the major modulating factor when either hopping at different frequencies (Farley, et al., 1999) or different contact periods (Hobara, et al., 2007). During single-leg hopping, vertical leg stiffness was greater at higher hopping frequencies (Austin, et al., 2002b) and similar to the value of vertical leg stiffness during DL hopping (Farley, et al., 1991; Farley & Gonzalez, 1996; Ferris, et al., 1997), with reduced ankle, knee and hip excursions. When hopping to a maximal height it was found that that knee stiffness was the primary determinant of leg stiffness and not ankle or hip stiffness (Hobara, et al., 2009). Differences in findings may be due to the method of assessment and differences in the tasks assessed. Hobara et al. (2009 & 2010a) assessed hopping at different frequencies and height while Farley and Morgenroth (1997) and Hobara et al. (2007) assessed hopping at a common frequency and different height and contact times. The interaction in mechanical behaviour between the ankle and knee in repeated landing tasks remains unclear and provides impetus for investigation during hopping due to the differences in testing protocols.

In comparing hopping to 'maximal height' to the preferred frequency, there were greater values for leg stiffness (~2 times), ankle stiffness (1.9 times) and knee stiffness (1.7 times) (Farley, et al., 1999). A computer model showed that changing one joint only by the same factor (1.9 and 1.7) changed leg stiffness only when ankle joint stiffness was changed and not when knee joint stiffness was adjusted (Farley, et al., 1999). Similarly, leg stiffness was shown to double when hopping on surfaces ranging in stiffness from soft to hard and ankle torsional stiffness increased by a factor of 1.75 (50% reduction in ankle excursion and 19% increase in moment) with reduced knee excursion (Farley, Houdijk, & Van Strien, 1998). In the study by Farley et al. (1998), participants landed with straighter knees on softer surfaces but the actual knee angle at initial contact increased from hard to soft surfaces. Furthermore, Farley et al. (1998) demonstrated that changing ankle stiffness only, in a computer model, increased leg stiffness by 1.75 and changing knee angle only, increased leg stiffness by 1.3, therefore the authors concluded that both joint stiffness and limb geometry account for changes in leg stiffness for maintenance of hopping on different surfaces. Farley et al. (1998) also reported that muscle activation levels and timing were not related to increasing ankle stiffness from hard to soft surfaces as EMG amplitude decreased and no change in timing was found (Farley, et al., 1998). Farley et al. (1998) suggest that rather than an increase in muscle activation amplitude of lower limb muscle muscles' characteristics, a change in the firing frequency of active motor units or recruitment of different motor units may be responsible for a change in muscle stiffness and therefore joint stiffness. While Farley et al. (1999) demonstrated that increasing hopping effort from a preferred steady state to a maximal effort may be controlled by the ankle, this conflicts from more recent study by Hobara et al. (2010) that did not show any change in ankle joint stiffness (Hobara, et al., 2010a). Evidence as to whether mechanical behaviour in the leg is discriminatory between ankle and knee joints remains conflicted.

The literature suggests that there is difficulty in comparing studies with different temporal and kinetic measures. Hopping at decreasing frequencies (3.0 down to 1.5Hz) resulted in increases in vertical height displacement in flight phase, minimal change in ground reaction force (GRF), greater leg compression in contact, greater contact times and lower leg stiffness (Hobara, et al., 2010a). Increasing hopping effort (preferred frequency to maximal height hopping) saw an increase in vertical height displacement in flight, greater GRF, lower leg compression in contact, lower contact time and greater leg stiffness (Farley, et al., 1999). The results for contact periods showed reverse trends with increasing vertical height displacement between hopping conditions. Hobara et al. (2010) compared hopping frequencies with participants instructed to maintain as short a contact time as possible (Hobara, et al., 2010a). Hobara et al. reported that a mean contact time of 0.18s (SEM) 0.02 at a hopping frequency of 3.0Hz that approximated the mean contact time during 'maximal height' hopping of 0.19s (0.01 (SEM)) in the previous study by Farley et al. (1999) (Farley, et al., 1999). Therefore, instructing the participants to hop with as 'short a contact time as possible' in the study comparing hopping frequencies by Hobara et al. (2010), may have modulated ankle joint stiffness. Thus, any change in hopping frequency had to be generated by a change in knee joint mechanical behaviour rather than the ankle joint, which contrasts the study by Farley et al. (1999) that identified the ankle joint as the primary joint for modulation of lower limb mechanical behaviour in hopping (Farley, et al., 1999).

Modulation of kinetics and kinematics is task-specific and highly dependent on the instructions and therefore the constraints placed on the participant (Arampatzis, et al., 2001; Cowling, et al., 2003). Hence, even when kinetic measures are similar, the motor control strategy may differ between participants. These findings, in healthy populations, of contrasting joint mechanical behaviour with the change in demand of hopping to greater heights (i.e.; hopping at 1.5Hz frequency or hopping to a maximal height) illustrate that the joint interaction is complex. Therefore, examination of these mechanical behaviours in a population of elite level volleyball and basketball athletes is necessary to identify strategies that may be specific to this sporting population that has a high prevalence of overuse knee injuries (Lian, et al., 2005).

Two of the studies so far cited report a modulation in ankle stiffness between hopping conditions related to changes in hopping frequency and surface stiffness (Farley, et al., 1998; Farley, et al., 1999) while a more recent study investigated changes in DL hopping at the same frequency (2.1Hz) with a preferred strategy and a 'stiff knee' strategy (Hobara, et al., 2007). In the 'stiff knee' condition the contact time was shorter and leg stiffness greater (~31%) with greater soleus and MG activity. Hobara et al. (2007) suggested that a change in ankle stiffness was the reason for change in leg stiffness, claiming that minimal activity of muscles other than at the ankle, would ‘contribute minimally’ (Hobara, et al., 2007). However, this is not evidenced by the recorded measures, as there was no EMG of the QF or hamstring muscles and no measure for ankle and knee joint moments. Instructing participants to maintain a ‘stiff knee’ does not predicate that there is a complete lack of movement at the knee and in a muscle with SEC such as the QF and during a high speed and movement such as hopping that generates a large GRF, there may be some knee movement. Indeed, the very instruction to actively hold the knee joint straight may activate the QF to a maximal capacity. Thus, it may be possible that there is a large knee moment being generated at the knee joint (large QF contraction effort) and if even a small amount of movement occurred there may be a high value of knee joint stiffness. During hopping it cannot be assumed that there is no change in proximal joints and as discussed so far, a number of independent variables may affect each joint in a different manner.

A large increase in GRF (~57%) and ankle joint moment (~52%) were reported between different cyclical loading tasks that were preferred and maximal height hopping (Farley, et al., 1999). However, both conditions had similar touchdown angles (35º to 41º plantarflexion at the ankle and 26º to 30º flexion at the knee) with significant reductions in ankle (ranging from 23º to 17º dorsiflexion) and knee joint excursions (ranging from 32º to 26º). The question arises as to how the ankle joint could produce such a large and significant increase in ankle joint moment when there was in fact a reduction in ankle joint excursion. This implicates increased stiffness of the eccentrically acting triceps surae MTU to resist the ankle being forced into flexion. Unfortunately, there was no EMG data collected in this study. However, it is clear that the mechanism for absorption of force at the ankle and knee joints remain unexplained.

The studies cited report that both ankle and knee joint function may modulate lower limb mechanical behaviour. However, it remains unclear as to how lower limb mechanical behaviour may adapt in elite jumping athletes with specific attention to the knee joint and the incidence of ‘jumper’s knee’ that is common in this group of athletes. Appreciating the contribution that each joint makes in absorbing the kinetic energy of the body as it lands in the braking phase is vital to furthering our understanding of both motor control and pathogenesis of injury when athletes are exposed to repeated landings.

2.2.2 Muscle Activation Characteristics during Hopping

The timing of muscle activation during the pre-loading phase, short latency stretch reflex and the functional stretch reflex are considered to contribute significantly to the preferred hopping frequency of 2.2Hz in humans (Figure 2.2.2). Economy of motion and conservation of energy are suggested as reasons for a ‘preferred’ steady state of flight and contact phase in gait. An inverse relationship between vertical stiffness (normalised) and aerobic demand in sub-maximal velocity ($3.35 m.s^{-1}$) running was reported for well-trained men (Heise & Martin, 1998) and supported by the finding of greater aerobic demand, with a deliberate running style of increased hip and knee flexion or ‘Groucho’ running (P. J. McMahon, Valiant, & Frederick, 1987). The period of muscle activity was hypothesised to optimise muscle energetics through the better utilisation of elastic energy from tendinous tissue (Allum & Mauritz, 1984). A short latency stretch reflex following landing, is observed to occur in the soleus and MG muscles ranging from ~37ms to ~50ms post-landing (Allum, et al., 1984; Duncan & McDonagh, 2000; Dyhre-Poulsen, Simonsen, & Voight, 1991; Funase, et al., 2001; Hobara, et al., 2007; Liddell & Sherrington, 1924; Moritani, Oddsson, & Thorstensson, 1991; Santello, 2005; P. Voight, Dyhre-Poulsen, & Simonsen, 1998) (figure 2.2.3). The duration of activity in both soleus and MG increased as duration of contact increased when the interval between hops ranged from 300 to 600ms (Funase, et al., 2001).

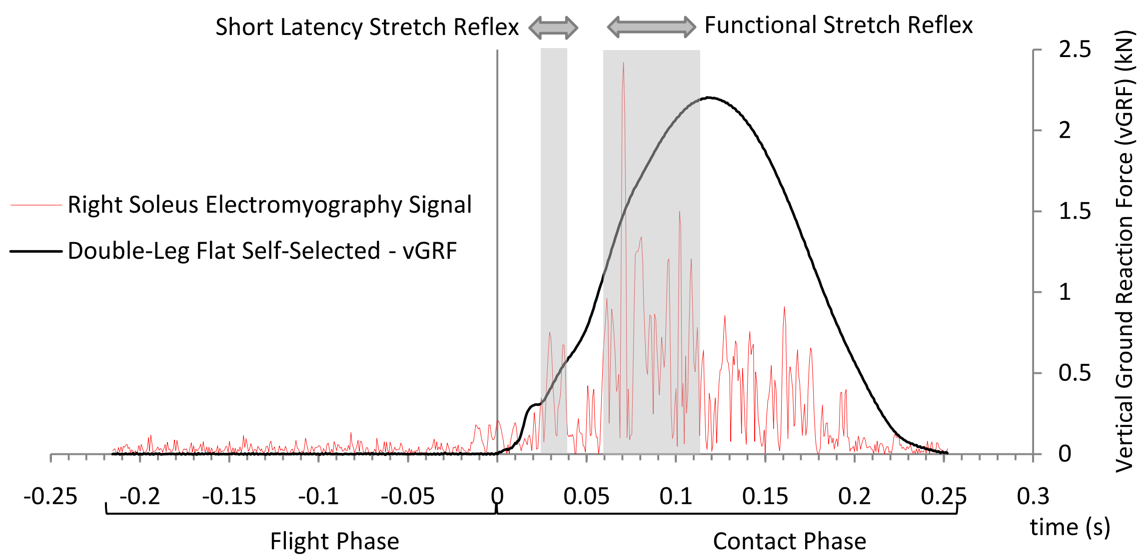


Figure 2.2.2 An electromyography (rectified and band-pass filtered 20 to 500Hz) recording of the right soleus muscle and vertical ground reaction force recording during double-leg hopping on a flat surface at a self-selected pace (2Hz) for a single hop (flight and contact phases). The short latency stretch reflex and the functional stretch reflex are evident in the braking period of the contact phase (shaded grey regions) (example from data collected in the current study).

The MG had a clear period of activation 100ms prior to loading and is described as a feed-forward mechanism that was consistent across hopping with varying intervals (Funase, et al., 2001; Gollhoffer & Kyröläinen, 1991) (Figure 2.2.3). This supported the previous findings during single-leg hopping of feed-forward or anticipatory activity in the MG, from 45ms (Moritani, et al., 1991) to 85ms prior to loading (Melvill-Jones, et al., 1971). The anticipatory mechanism is one made without sensory feedback in contrast to feedback mechanisms that are made in response to sensory feedback (Seidler, Noll, & Thiers, 2004) as exampled by the short latency stretch reflex. Soleus muscle activity observed prior to landing, when there were short intervals between hops, was attributed to preparation for the propulsive phase, rather than preparatory for the loading phase, as it was similar to the activity observed in a take-off only task (no hopping). Tibialis anterior (TA) had a reverse pattern of activation that was high in flight phase and ended before initial contact and remained low in contact phase (Dyhre-Poulsen, et al., 1991; Hobara, et al., 2007).

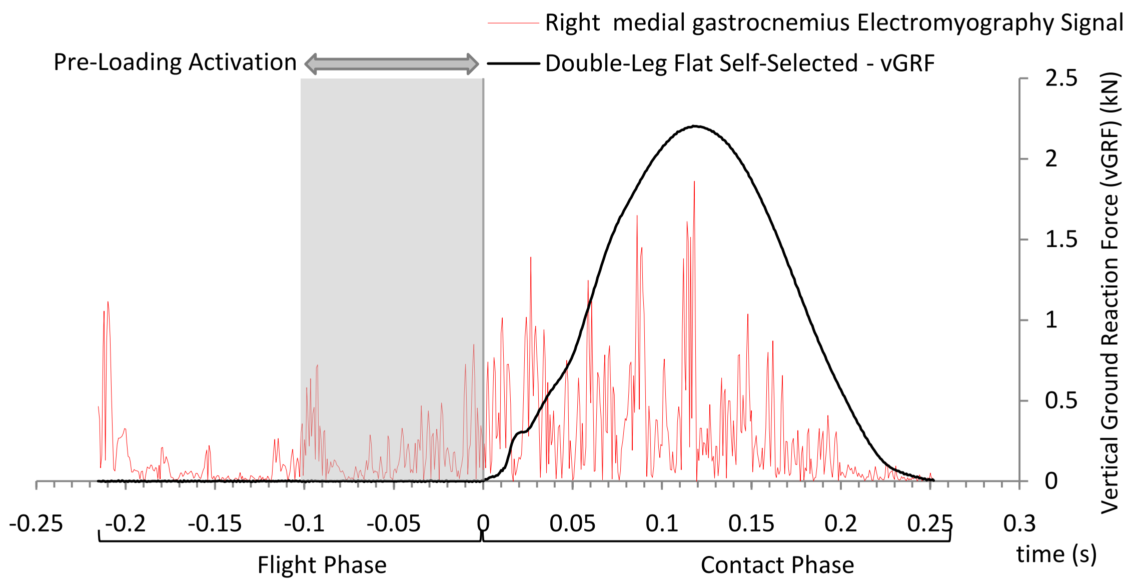


Figure 2.2.3 An electromyography (rectified and band-pass filtered 20 to 500Hz) recording of the right medial gastrocnemius muscle and vertical ground reaction force recording during double-leg hopping on a flat surface at a self-selected pace (2Hz) for a single hop (flight and contact phases). The pre-loading activation period is evident in the flight phase (shaded grey region 100ms prior to landing) (example from data collected in the current study).

The EMG amplitude of the short latency reflex and the functional stretch reflex that describes EMG activity for propulsion in the subsequent hop appear optimised at a hopping frequency of 2Hz (hopping interval 460ms) (Funase, et al., 2001; Melvill-Jones, et al., 1971). Importantly, the pre-loading activation amplitude of MG was demonstrated to increase when hopping with greater leg stiffness compared to hopping with a self-selected style at the same frequency (2.1Hz) (Hobara, et al., 2007). The finding of greater leg stiffness during the 'knee stiff' condition was suggested to be due to an increase in ankle torsional stiffness although it was not measured (Hobara, et al., 2007). This demonstrated that the pre-loading activation amplitude of the MG is subject to change in amplitude, based on the hopping characteristics primarily that of the time before landing is anticipated and the height from which the body is falling (Santello, 2005). In overground running over obstacles of different height, MG pre-loading activation decreased as the height of the step increased and was attributed to the demand of changing leg posture for a change in step height (Müller, Grimmer, & Blickhan, 2010). Müller et al. (2010) found no correlation between MG pre-activation and ankle stiffness in these tasks. Thus, it has been demonstrated that leg stiffness does change concomitant to changes in MG muscle activity (Hobara, et al., 2007) and that there are differences in leg stiffness between participants with different levels and types of training background (Hobara, et al., 2008; Hobara, Kimura, Omuro, Gomi, Muraoka, Sakamoto, & Kanosue, 2010b). It may be expected that under different hopping conditions, that mechanical and muscle activation characteristics may be specific to athletes who participate in high intensity loading activities such as volleyball and basketball. Therefore, the finding that the MG feed-forward mechanism changes during rapid loading activities warrants investigation in elite level athletes during hopping.

The reason for modulation of the anticipatory activity in MG is unclear. MG pre-loading activity is shown to change in running over different obstacle courses (Müller, et al., 2010). However, is the altered MTU stiffness a functional requirement or are the recordings of different MG activation amplitudes a result of joint angle differences (ankle and/or knee) in landing? If there is a strong correlation of MG anticipatory amplitude activity based on joint angles then findings of modulation of this activity may not be a true representation of the altered MTU stiffness prior to landing, rather a global strategy by the nervous system to attempt to stiffen the triceps surae complex prior to landing regardless of the magnitude of the GRF. It is suggested that stretch reflexes contribute (Sinkjaer, Toft, Andreassen, & Hornermann, 1988; Trimble, Kukulka, & Thomas, 2000) but are not the major controller to muscle activity observed following landings (Santello, 2005) and that there is a central program that drives the voluntary activity seen during high velocity actions such as during hopping (Duncan, et al., 2000; McDonagh & Duncan, 2002).

2.2.3 Hopping in Elite Athletes

Differences between different sporting groups may be attributed to their training history and the self-selection process of specific physical attributes being well suited to the sport (Bompa, 1999; Gabbett & Georgieff, 2007; Mohamed, Vaeyens, Matthys, Multael, Lefevere, Lenior, & Philppaerts, 2009; Reilly, Morris, & Whyte, 2009; Sheppard, Gabbett, & Stanganelli, 2009). Differences in mechanical behaviour were detected between power (sprint running) trained athletes compared to endurance trained athletes, with power athletes having greater leg stiffness across both hopping frequencies (1.5 and 3.0Hz); however at the lower frequency they had greater knee joint and lower ankle joint stiffness (Hobara, et al., 2008). EMG activation patterns of lower limb muscles were similar for both groups.

Endurance trained athletes had greater ankle and knee joint stiffness than untrained participants, while at a common hopping frequency (2.1Hz) there were shorter contact times and greater leg stiffness in the athletic group (Hobara, et al., 2010b). An important note in this previous study was that there was approximately 30% greater peak moment in athletes than the untrained participants with a mean of 4.35Nm/Body Mass (SD) (0.45) compared to 3.16Nm/Body Mass (0.67). However, statistical significance was not detected due to the study being under powered and the variance in the untrained population being large. Hobara et al. (2010) suggested that differences in connective tissue properties of the calf complex between trained and untrained groups may explain the difference in ankle torsional stiffness. However, it is clear that a greater ankle peak joint moment contributed to greater ankle joint stiffness in the athletes. Knee joint moments were similar between groups while knee angular excursions were lower in athletes (Hobara, et al., 2010b). Thus, it is apparent that knee joint stiffness was different due to differences in angular excursion while ankle joint stiffness was different due to a combination of differences in angular displacement and joint moment. The key finding by Hobara et al. was that ankle and knee joint stiffness at sub-maximal and preferred hopping frequencies are dependent on the training history (Hobara, et al., 2010b). It is unclear if such differences in strategy would exist in elite athletes with different training histories and provides the basis for evaluation of elite jumping athletes.

The research into hopping demonstrates that motor control strategy is adapted to the specific hopping conditions. Differences in motor control strategy are also dependant on the training history of individuals. Therefore, hopping is a motor skill that yields itself to being assessed in both the clinical and laboratory settings to identify changes in lower limb mechanical behaviour during the SSC. This is of interest in this thesis that aims to assess the lower limb mechanical behaviour in elite jumping athletes that expose themselves to high loads in the lower limb.

2.3 Research Goals and Aims

The purpose of this chapter was to develop a protocol to assess lower limb motor control in elite jumping athletes including lower limb joint mechanical behaviour and muscle activation characteristics.

Key research aims were to:

- evaluate the knee and ankle interplay during hopping in elite jumping athletes.

3 Chapter Three: Lower Limb Motor Control in Elite Athletes during Hopping

3.1 Chapter Aims

This chapter aims to:

- develop a hopping protocol to assess kinetic, kinematic and muscle activation characteristics for elite jumping athletes.
- evaluate kinetic, joint and muscle activation changes during a series of hopping tasks for elite jumping athletes.
- evaluate the interaction between the knee and ankle in measured variables during a series of hopping tasks for elite jumping athletes.
- evaluate the inter-side differences in kinetic, joint and muscle activation amplitude variables during a series of hopping tasks for elite jumping athletes.

3.2 Introduction

Evaluating motor control parameters during repeated SSC loading tasks would be ideal in their natural environment during competition or training. However, this is not always feasible due to technological limitations. Hence, a protocol utilising the skill of hopping was developed to assess lower limb motor control during hopping in elite jumping athletes who are susceptible to patellar tendinopathy (Lian, et al., 2005).

The role of the knee and its interplay with the ankle was the focus of this study. The PT is the primary connective tissue responsible for a transfer of force from the QF muscle to the tibia with bony attachments from the inferior pole of the patella to the tibial tuberosity (Waligora, Johanson, & Hirsch, 2009). The anterior fibres of the PT are contiguous with the superficial fibres of the rectus femoris over the patella (Waligora, et al., 2009; Wangwinyuvirat, Dirim, Pastore, Pretterklieber, Frank, Haghighi, & Resnick, 2009). To help understand the pathogenesis of patellar tendinopathy in jumping athletes it is vital to gain an understanding of how the knee functions during landing in elite jumping athletes.

A number of factors were considered in using hopping as a model to evaluate the role of the knee in elite jumping athletes. Hopping is a repeatable and rhythmical motion (Zehr, 2005).

A key reason to employ repeated loading of the lower limb is that damage to tendon tissue is most profound when the load is oscillating (Simonsen, Kligaard, & Bøjsen-Møller, 1995; Wang, Ker, & Alexander, 1995; Wren, Lindsey, Beaupré, & Carter, 2003). Wang et al. (1995) demonstrated that applying an oscillating load at low and high frequencies both resulted in damage to failure of wallaby tendon tail sooner than a constant load (Wang, et al., 1995). The aim of the current study was not to damage tendon tissue during hopping, rather to evaluate the behaviour of the motor control system when the knee was exposed to repeated or oscillating loading at the knee. The factors of oscillating the load and a limited capacity for repeated storage and utilisation of elastic energy of the PT may be specific to the pathogenesis of patellar tendinopathy (D. P. Richards, Ajemain, Wiley, & Zernicke, 1996). An assessment of the lower limbs and their mechanical behaviour may shed light on some of the factors that change under different loading conditions.

Manipulation of the skill of hopping was aimed at changing the demand on the sensory-motor system to mimic sports, where demand on the sensory-motor system changes between motor tasks. A number of motor control theories are proposed to describe how the motor system enables this control and change in its behaviour. Some of the common theories are the 'Information Processing Approach' (Schmidt & Lee, 2005), 'Dynamical Systems Approach' (van Emmerik & van Wegen, 2000), 'Computational Approach' (Wolpert, 1997) and 'Equilibrium Point Theory' (M. Latash, 2010; M. L. Latash, 2008).

Rather than critique the different motor control theories, this investigation assumes that the motor system needs to modify a single strategy or change between strategies when the goal of the motor skill changes. To this end, the focus of this investigation was to evaluate the change in lower limb mechanical behaviour during hopping, when the three factors of hopping pace, surface and task were manipulated. It may be debated whether the athlete followed the principle of optimisation (M. L. Latash, Scholz, & Schoner, 2002; Mulder, Zijlstra, & Geurts, 2002), principle of abundance (Gelfand & Latash, 1998) or another principle that seeks to describe how the motor system changes. A key focus was to measure kinetic, kinematic and muscle activation that describe the 'output' to function that is controlled by the nervous system centrally and peripherally.

While motion capture allows analysis of kinematics, MTU stiffness is not as easily determined. There is no gold standard means to measure tensile force or strain for a MTU *in vivo* during a high speed and high force activity such as hopping (Fleming & Beynnon, 2004). Direct measures of tendon strain and force have been made using implantable devices (liquid metal strain gauge, buckle, force and pressure transducers and optic fibre insertions) however, these methods have inherent draw backs as they require surgical implantation, may change the behaviour of the tissue due to the insertion and are often difficult to calibrate accurately (Fleming, et al., 2004; Herzog, Hasler, & Leonard, 1996; Ravary, Pourcelot, Bortolussi, Konieczka, & Crevier-Denoix, 2004). These procedures may cause pain and alter the performance of the tasks being tested (Fleming, et al., 2004; Ravary, et al., 2004). Furthermore, these invasive methods are not able to accurately quantify the relative force contribution of the contractile and non-contractile components (series and parallel elastic components). The non-contractile elements to able to store and release elastic energy which contribute to force generation by the MTU during dynamic tasks.

A number of derived variables including vertical leg stiffness, contact time, joint stiffness, joint excursion and muscle activation level via surface electromyography (sEMG) have been utilised to describe motor behaviour during various SSC tasks. Many of these variables have been shown to be associated with performance outputs, for example, foot posture during standing and running was correlated with leg stiffness in running (Viale, Dalleau, Freychat, Lacour, & Belli, 1998) and vertical leg stiffness is also reported to have an association with the acceleration and deceleration phases in elite level 100m sprinters (Bret, Rahamani, Dufour, Messonnier, & Lacour, 2002). In contrast, an earlier study reported that vertical leg stiffness was correlated with maximal velocity and not with acceleration (Chelly & Denis, 2001). Consistent with these experimental protocols, the current study evaluated kinetic, kinematic and muscle activation variables during a series of steady state sub-maximal hopping SSC tasks in elite level jumping athletes. Joint variables in the current study refer to motion in the saggital plane.

3.2.1 Hopping Conditions

In the current study, hopping was performed at both a self-selected and forced pace. By controlling the hopping frequency during the forced pace condition, any change in lower limb motor control strategy could be attributed to the other two factors that were manipulated. These two factors were the type of hopping task and the angle of support surface on which the individual was hopping.

Double-Leg and Alternate-Leg Hopping

Hopping tasks included DL hopping, that is, when both lower limbs act synchronously and alternate-leg (AL) hopping, that is, when the contact phase alternates between limbs. The terms DL and AL hopping are used throughout this study, as is the standard term in this area of research. The term jumping refers to a single effort of propulsion, flight and landing and is distinguished from hopping that is a repeated cycle (flight and contact). The majority of studies evaluating hopping have examined healthy participants and used DL hopping, with few studies using a single-leg task (Austin, et al., 2002a; Austin, et al., 2002b, 2003; Chang, et al., 2008; Lichtwark, et al., 2005b; Melvill-Jones, et al., 1971; van Uden, Bloo, Kooloos, van Kampen, de Witte, & Wagenaar, 2003). Single-leg loading strategies were of interest in this investigation especially as a study of women's volleyball identified 45% of offensive and 43% of defensive landings to be single-leg (Tillman, Hass, Brunt, & Bennett, 2004). A single-leg loading strategy is also prevalent in human gait during walking, running and landing (Austin, et al., 2002b). Alternate-leg hopping, or on-the-spot jogging, more closely mimics functional activities such as running and single-leg landing. Single-leg landing is rarely performed with repeated loading of the one limb and more often performed with legs contacting the ground in series.

In volleyball it was demonstrated that there are two specific styles of landing including DL and a step-close technique when there is a temporal delay between the two landing legs (Coutts, 1982; Gutiérrez-Davilla, Campos, & Navarro, 2009). The latter technique is shown to reduce the force in landing and the choice of jump-land strategy is attributed to the state of play in the field sports and the interaction of the athletes with physical constraints of opposition players or rules. The step-close technique may be the strategy of choice if there is a requirement for a shorter contact period to optimise ball contact or timing of jumping, or to reduce the horizontal velocity at take-off to avoid potential contact with the net in volleyball (Coutts, 1982). A step-close landing strategy may also share the load between legs such that the first leg in contact may be more conservative in its mechanical behaviour with the expectation that the contralateral leg will also be used, thus optimising landing mechanics for the individual. Therefore, differences in landing strategy during braking between AL compared to DL hopping, may be due to an inherently different strategy being utilised or due to a an optimisation of motor control strategies available to reduce the risk of over load of either leg.

Asymmetry between dominant and non-dominant legs has been shown to be task dependant with Flanagan et al. (2007) reporting that landing from a single-leg rebound jump (3 consecutive single-leg hops) and drop jump had no asymmetry in vertical leg stiffness between the dominant and non-dominant legs and that the non-dominant leg was stiffer during the rebound jump task (Flanagan & Harrison, 2007). A computer model detected no difference in vertical jump height during a counter-movement jump when comparing models with and without leg strength differences (10% difference in physiological cross-sectional area of lower limb muscles) between sides (S. Yoshioka, Nagano, Hay, & Fukashiro, 2010). These findings demonstrate that the mechanical behaviour may be different for each leg and any difference is task-specific and independent of the measure of the performance. The current study aimed to identify differences between legs in mechanical behaviour in both DL and single-leg loading strategies during unsupported weight-bearing, using a repeated SSC across different conditions.

A number of factors are likely to have contributed to the use of DL hopping in previous studies. These include an assumption that kinetic, kinematic and muscle activation are similar between sides and that there is less demand on balance, strength and coordination during DL compared to AL hopping. A technological limitation that also supports the use of DL hopping strategy is that two force plates are required to identify individual leg loading profiles. When a single force plate is used the recorded GRF is the result of both legs acting in combination (Padua, et al., 2006) and does not allow for an evaluation of inter-side differences in leg or joint mechanical behaviour. In the current study some of these issues are addressed with a method utilising dual force plates to measure kinetic and kinematic variables independently for each leg along with EMG for each leg.

Hopping is physiologically constrained by limitations on the length of joint lever arms (mechanical) and the available energy in the form of adenosine triphosphate at the site of muscular contraction, primarily utilised in concentric muscle action (Caruso, Hernandez, Saito, Cho, & Nelson, 2003; Dudley, Tesch, Harris, Golden, & Buchanan, 1991). In the current study, AL hopping, or on-the-spot jogging, was performed in addition to DL hopping. Alternate-leg hopping was introduced to overcome some of the confounding variables associated with single-leg hopping that include fatigue, substantial variations of the upper body segments and translatory (lateral or in the anterior posterior planes) displacement of the contact points on the force plate. There was an assumption that AL hopping had a reduced translatory motion at the point of contact on the force plate (x and y axes) compared to single-leg hopping that had the potential to have an inconsistent landing position. During hopping, the line of gravity passes through the base of support and the individual will try and maintain a similar spatial position for landing by moving across the transverse plane (Figure 3.2.1). A pattern of greater deficits in balance were noted when single-leg landing from a sub-maximal effort compared to DL jumps in either the lateral or diagonal directions (Wikstrom, Tillman, Schenker, & Boras, 2008). It is hypothesised that there are more profound compensatory strategies used in single-leg hopping compared to either DL or AL hopping. Although this has not been investigated specifically, AL rather than single-leg hopping was chosen for assessment of a single leg-loading strategy to reduce the possibility of detecting changes in variables due to compensations which may be a result of correcting the landing position.

The individual may select strategies to adjust both contact and flight phases to alter or maintain the hopping frequency (quotient of hop cycles to time). Flight time may be altered in two ways. First, flight time can be altered by changing the vertical height of the hop and second, by developing a strategy of flexing or extending the lower limb during flight period to modulate the contact on the force plate. The individual may alter contact phase once again by changing the degree of lower limb flexion in flight time and also minimise the delay of the contact period on the force plate. It is impossible to have no vertical height displacement (in flight) during hopping otherwise the individual will constantly be in contact. The individual may choose to co-adapt both contact and flight phases to maintain hopping frequency. During DL hopping, hopping frequency may be influenced by both legs, however in AL hopping, hopping frequency is dependent on the single leg.

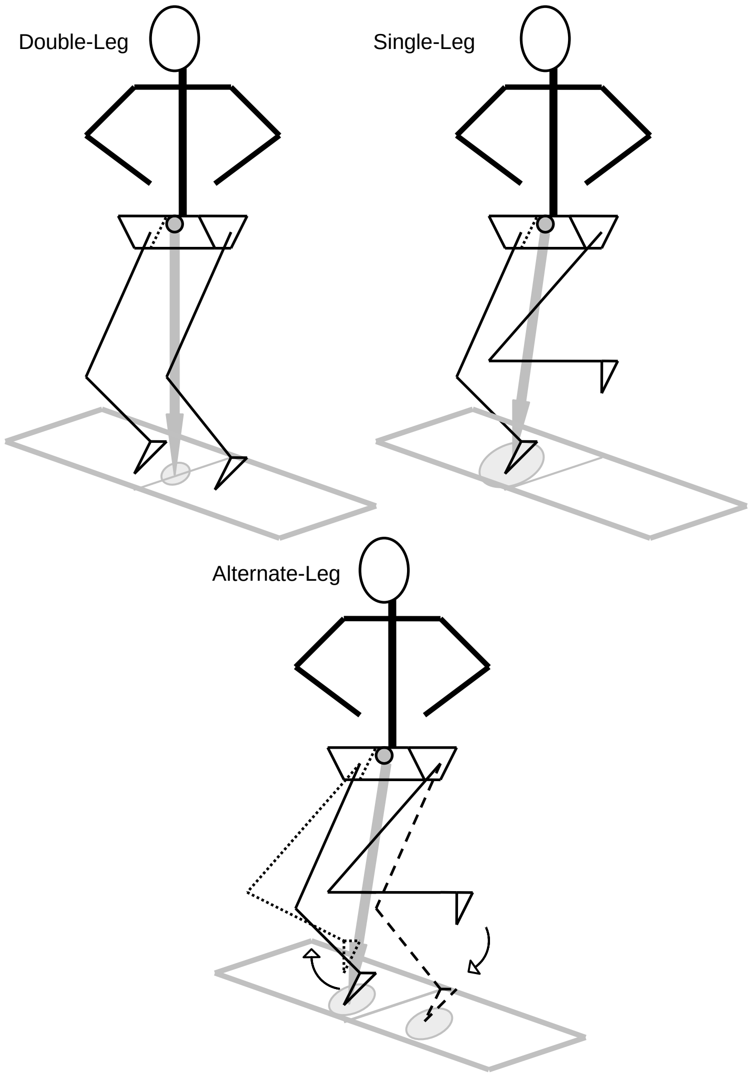


Figure 3.2.1 Schematic representation of double-leg, single-leg (SL) and alternate-leg (AL) hopping on dual force plates (grey rectangles) under each leg. The centre of mass (grey shaded circle) generates a moment about the contact point through which the line of gravity (shaded grey arrow) passes. The relative areas of where the base of support will be are shown (shaded grey ellipsoid). Note that this area is likely to be less during AL than SL hopping when the left leg contacts the ground (large dashed lines) and the right leg leaves the ground (dotted lines).

Hopping Support Surface

Hopping was performed on both a flat and decline surface. The basis for performing hopping on a decline surface stems from the use of the single- leg decline squat that was originally introduced as an assessment tool to discriminate the tensile structures, specifically the PT, during a functional loading manoeuvre (Purdam, Cook, Hopper, Khan, & VIS tendon study group, 2003). The finding that this manoeuvre elicited significantly greater pain response at the site of injury in athletes with patellar tendinopathy compared to other manoeuvres, suggested that it may more specifically load the PT (Purdam, et al., 2003). Following this finding, the use of a decline board has been used in rehabilitation exercise programs to preferentially load the PT (Kongsgaard, Aagaard, Roikjaer, Olsen, Jensen, Langberg, & Magnusson, 2006; Purdam, et al., 2003; Purdam, Jonsson, Alfredson, Lorentzon, Cook, & Khan, 2004). Zwerver et al. (2007) demonstrated 40% greater knee moment performing a single-leg squat on a decline surface greater than 15º (Zwerver, Bredeweg, & Hof, 2007). This was primarily attributed to the COM moving posteriorly, as the angle of decline surface increased. Considering that the inertial components to the manoeuvre were reported as negligible, the primary reason for increased knee moment were attributed to a greater knee moment arm length from the line of gravity as decline angle increased.

Greater EMG amplitude of the QF and greater PT strain during single-leg 25º decline squat compared to single-leg squat on a flat surface, suggests greater loading of the knee extensor MTU using a decline surface (Kongsgaard, et al., 2006). Considering that the force-length relationship of the PT is curvilinear, this suggests that greater tensile force was being generated at the QF when squatting on a decline surface as this will correspond to greater PT strain. Kongsgaard et al. (2006) did not measure joint moments, however, the authors suggest that the COM moved posteriorly when on the decline surface compared to the flat surface, based on start and stop angles at the ankle and hip, while the knee flexion angle was controlled at 90º (Kongsgaard, et al., 2006).

A reduction in ankle moment (>32%) at decline angles of 16º and 24º was demonstrated with an increase in knee moment (>21%) compared to single-leg decline squat on a flat surface (J. G. Richards, Thewlis, Selfe, Cunningham, & Hayes, 2008). Concomitant to these changes were increased QF and gastrocnemius muscle activity with greater decline angles (Kongsgaard, et al., 2006; J. G. Richards, et al., 2008). Richards et al. (2008) suggest that the increase in gastrocnemius muscle activity may be due to the increase in demand as a stabiliser and that it works with TA to co-contract at the ankle joint (J. G. Richards, et al., 2008). However, this mechanism for ankle stabilisation is unsubstantiated as EMG activity of the TA was not collected.

Frohm et al. (2007) had similar results examining differences between flat and decline surfaces during single-leg squat with weighted exercises, finding 25 to 30% greater PT force on the decline surface (Frohm, Halvorsen, & Thorstensson, 2007a). The absolute values calculated for PT force (determined from dividing knee moment by PT moment arm length) may be imprecise, as PT moment arm length values were used from a previous investigation (Smidt, 1973) even though the authors state that this measure is 'subject specific' and the populations across the two studies varied in height (mean (SD)) (176cm v 185cm (5)). However, the repeated measures design of the current study does not detract from the previous findings that reported greater mechanical work at the knee (30%) and increased QF and MG mean EMG amplitude on the decline (25º) compared to flat surface (Frohm, et al., 2007a). The evidence suggests that during the single–leg decline squat the decline surface affords changes in kinetics and kinematics between the knee and ankle joints with a bias for greater moments at the knee joint and reduced moments at the ankle joint. No study has investigated these parameters during a SSC task performed on both flat and decline surfaces, providing further impetus for an investigation in elite jumping athletes who are susceptible to injury at the knee joint.

3.2.2 Knee and Ankle Interplay

As discussed in chapter 2, the conditions under which hopping is being tested govern the ability of either the ankle or knee joint to modulate lower limb mechanical behaviour. The preference to load the left or right side asymmetrically may be a risk factor for the pathogenesis of patellar tendinopathy. Examination of 10 elite level male volleyball players distinguished the feature of greater knee joint excursion during landing from a spike jump as highly predictive of left sided patellar tendinitis (D. P. Richards, et al., 1996). Richards et al. correctly resist the temptation to see this as a causative factor for patellar tendinopathy as the study used a cross-sectional design (D. P. Richards, et al., 1996). The concept of an asymmetry in landing mechanics is strengthened by a recent finding that landing from a spike jump, under two different take-off to land distances (normal and long), detected a tendency for the centre of pressure to be directed under the left more than right side (Marquez, Masumura, & Ae, 2009). Bearing this in mind, the current study aimed to quantify any inter-side asymmetries in cyclical steady-state loading of the lower limbs.

In a follow-up study on elite volleyball players Richards et al. (2002) detected that high ankle inversion moments were associated with patellar tendinopathy (D. P. Richards, Ajemain, Wiley, Brunet, & Zernicke, 2002). Furthermore, high knee and ankle joint loading rates were demonstrated in volleyball players with a previous history of patellar tendinopathy compared to those with a recent history or no history of patellar tendinopathy (Bisseling, Hof, Bredeweg, Zwerver, & Mulder, 2007, 2008). Bisseling et al. (2007 & 2008) detected that only peak knee moments were lower in those with a recent history of patellar tendinopathy than those with no history (Bisseling, et al., 2007, 2008) and that there was reduced knee flexion in the eccentric phase of landing from a spike jump (D. P. Richards, et al., 2002). These findings demonstrate that mechanical behaviour was associated with pain at the knee extensor mechanism. This necessitates a protocol to try and differentiate the change in mechanical behaviour at each joint in a population of elite level jumping athletes that have a high prevalence of overuse knee injuries (Lian, et al., 2005).

3.2.3 Muscle Activation Characteristics during Hopping

Muscle activation characteristics, specifically the pre-loading activation amplitude are shown to change during rapid loading activities such as hopping and running (Müller, et al., 2010; Santello, 2005). Observations that there is a modulation of the feed-forward mechanism in MG provided impetus for investigation when comparing the performance of the SSC across hopping conditions that may involve a change in leg stiffness. It remains unclear to the exact mechanism of how MG pre-loading activation contributes to energy absorption in the braking phase of landing. There is a strong suggestion that by activating the MG there is a stiffening of the MTU (contractile and non-contractile elements) that will benefit the resistance to ankle joint flexion (Santello, 2005). At the MG this is proposed to occur by the shortening of the contractile element that loads the Achilles tendon (non-contractile element) and with either maintenance of the ankle at a fixed angle or with increased dorsiflexion, there is further loading of the Achilles tendon in the opposite direction to the contraction of the contractile element of the MG. This cannot be confirmed until measures of MTU stiffness can be measured accurately during dynamic function. Although it is suggested that the pre-activation of the muscles significantly alters the lower limb stiffness, the literature suggests that a central motor control strategy drives the voluntary activity of the lower limb muscles during hopping (Duncan, et al., 2000; McDonagh, et al., 2002).

Training history has been shown to affect lower limb mechanical behaviour during hopping tasks. Hobara et al. (2008) reported greater leg stiffness in power compared to endurance athletes (Hobara, et al., 2008) and greater ankle and knee joint stiffness in endurance athletes compared to untrained participants (Hobara, et al., 2010b). The level of expertise of jumping in men also influences hopping and landing motor behaviour. Assessment of common activities such as running and jumping found a similar range of errors of range in performance for both experts and novices. Similarly common patterns of a change in leg stiffness with a change in jump height target have also been observed (Laffaye, Bardy, & Durey, 2005). However, expert jumpers, particularly volleyball players, had lower leg stiffness and GRF values, with greater time in contact during the jump than novices and other jumping athletes. In contrast to the reduced contact times in trained participants (Hobara, et al., 2010b), Laffaye et al. (2005) showed that expert jumpers appeared to increase the time in contact before take-off during a run and jump task (Laffaye, et al., 2005). Two factors may discriminate the motor control strategy used in loading tasks. First, exposure to sport-specific training and second, the motor skill assessed. Each may each lead an individual to use a specific motor control strategy or set of strategies. Therefore, to identify variables that would allow a detection of differences in lower limb loading and motor control strategies, examining a homogenous elite cohort under different controlled test hopping tasks may provide insight into the interactions between the sides and joints. A repeated measures design for hopping served as a tool to discriminate motor control strategies in elite level jumping athletes where there were similar lower limb loading histories.

Therefore, the purpose of this investigation was to evaluate the change in kinetic, kinematic and muscle activation characteristics in elite level jumping athletes during lower limb loading tasks. Of specific interest in this study were the characteristics of the knee joint and its interplay with the ankle, as elite level jumping athletes are susceptible to injury at the site of the knee extensor mechanism.

3.2.4 Hypotheses

The derived variables for each of the domains of kinetics, joint mechanics and muscle activation and the levels of each of the hopping conditions are described in the methods section (3.3.6, Table 3.3.1). This investigation tested the following hypotheses for differences in hopping frequency; kinetics, joint mechanics and muscle activation when hopping under 8 different conditions controlled by the factors of hopping frequency, task and surface.

1. That there is no difference in hopping frequency, kinetics, joint mechanics and muscle activation during hopping under different conditions.
2. That there is no difference between left and right sides in hopping frequency, kinetics, joint mechanics and muscle activation during hopping under different conditions.

3.3 Methods

3.3.1 Participants

Nine elite male athletes with a mean age of 18.4 years (SD) (1.5), 200.2cm (5.2) in height and 90.8kg (6.5) in weight were recruited for this study. The current investigation defined elite athletes as those who trained on a full-time basis (approximately 30 hours/week) to compete in their chosen sport. All participants were full time athletes in the volleyball or basketball programs at the Australian Institute of Sport, Canberra, Australia. All participants were free from knee pain at the time of this study. Participants were excluded if they had a current history of pain that may have been aggravated by the testing procedure(s). Participants were excluded if they reported any known medical or physical condition that may prohibit them from physical exertion, activity or any condition that may be aggravated by the testing procedure(s) (e.g. epilepsy, anterior cruciate ligament reconstructive surgery, sero-negative disorders, etc.).

This investigation was approved by the La Trobe University Faculty Human Research Ethics Committee (Appendix A) and the Australian Institute of Sport Research Ethics Committee (Appendix B). All volunteer participants provided signed informed consent (Appendix C). If participants were under 18 years of age, parental/guardian written consent was also obtained. All participants were able to withdraw their consent at any time during testing.

3.3.2 Equipment & Instrumentation

The equipment included an EMG system (Myosystem, Noraxon Inc. Az, USA) to allow recording of muscle activity of the left and right lower limbs muscles, a motion analysis system (VICON motion systems, Oxford, UK) to allow recording of kinematics and two multi-component force plates (multi-component force plate, 9287BA, Kistler Instrument

Corp., NY, USA) to allow recording of temporal and kinetic data (Figure 3.3.1). All data was collected using VICON Nexus software (version 1.3) from the EMG system, motion analysis system and force plates. The specifications of each system are described in Appendix D.

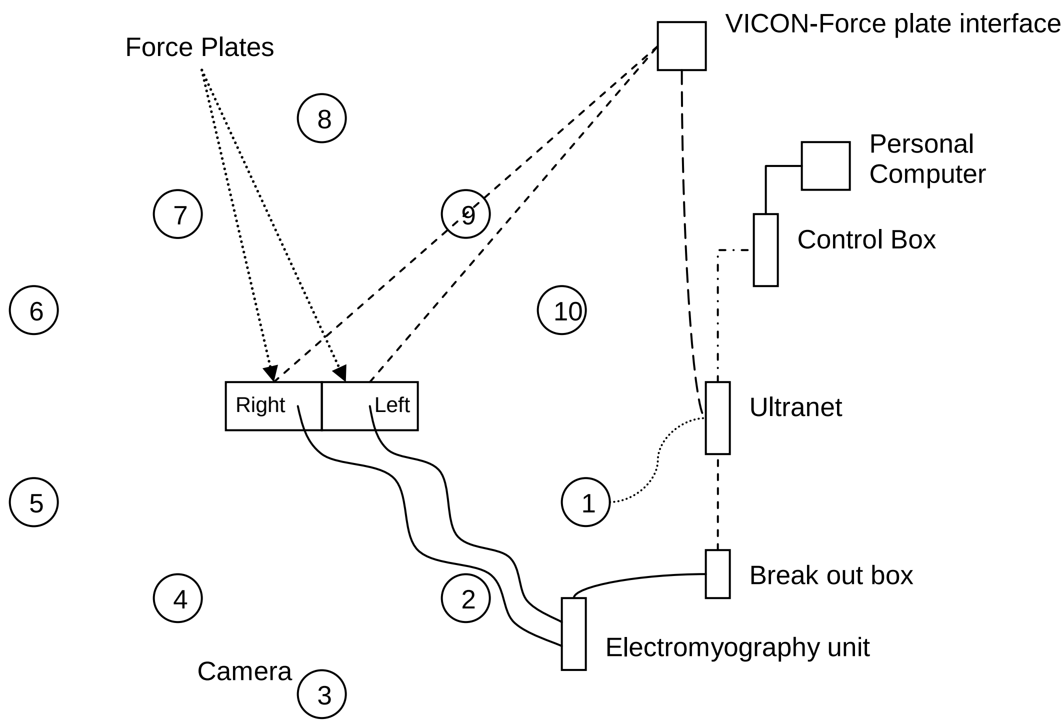


Figure 3.3.1 Schematic representation (aerial aspect) of the configuration of instruments with 10 cameras (numbered circles), two force plates (Left and Right), electromyography unit (EMG), cable from cameras to ultranet unit (shown for camera 1 only), cable from EMG unit to break out box (curved solid line), cable from break out box to ultranet unit (dashed line), cable from ultranet to control box (dashed and dotted line), cable from the VICON-force plate interface to ultranet (long dashed line) and cable from ultranet to personal computer (solid straight lines). The two sinusoidal lines represent the left and right active cables connecting the EMG unit to the pre-amplified leads over each muscle on the left and right legs of the participant.

Ten cameras (MX40 and MX13) were placed concentrically around the force plates to allow capture of markers placed on the participant (Figure 3.3.1). This was termed the capture volume. Each camera was aimed either towards the lower limbs, trunk or head to ensure the capture of all light retro-reflective markers in both static and dynamic trials. Each camera was connected to an ultranet unit with a fixed cable (shown only for ‘camera 1’ in Figure 3.3.1). The force plates were connected to the ultranet via the VICON-force plate interface (Figure 3.3.1) for collection of the force plate signals.

Surface EMG electrodes were each connected to a pre-amplifier and in turn connected to an active cable. The ‘active cable’ defines the presence of an isolated low voltage and low current present in the cable. Each active cable carried 6 sEMG leads from each leg. These two active cables (one each for left and right legs) in turn terminated in a 20 pin connector that connected to the Myosystem 1400A. A custom-made break-out box carried the analogue signal from the Myosystem 1400A to the ultranet unit.

Once the motion analysis, force plate and EMG systems were connected to the ultranet unit, this was connected to the MX control box that was in turn connected to a personal desktop computer where VICON Nexus software (version 1.3) allowed capture during each hopping trial. Surface EMG and force plate signal were sampled at 1500 Hz and motion capture was sampled at 250Hz.

Prior to testing it was ensured that all markers were visualised in the capture volume and that the force plate signals from each of the two force plates and EMG signals from the 12 muscles were being detected during static and dynamic tasks. This check was performed for each participant prior to a warm-up and familiarisation period and prior to data collection for each trial with all systems calibrated via the VICON software interface.

Two decline boards were constructed from solid timber and covered with non-slip rubber matting on the decline surface that was at a 23º angle to the horizontal. A schematic representation of the set up with a participant standing on decline boards is shown in Figure 3.3.2. There was a 2mm gap between adjacent force plates. Each decline board was aligned on the left and right force plate respectively, fitting inside the boundaries of the force plates, while ensuring a gap was maintained.

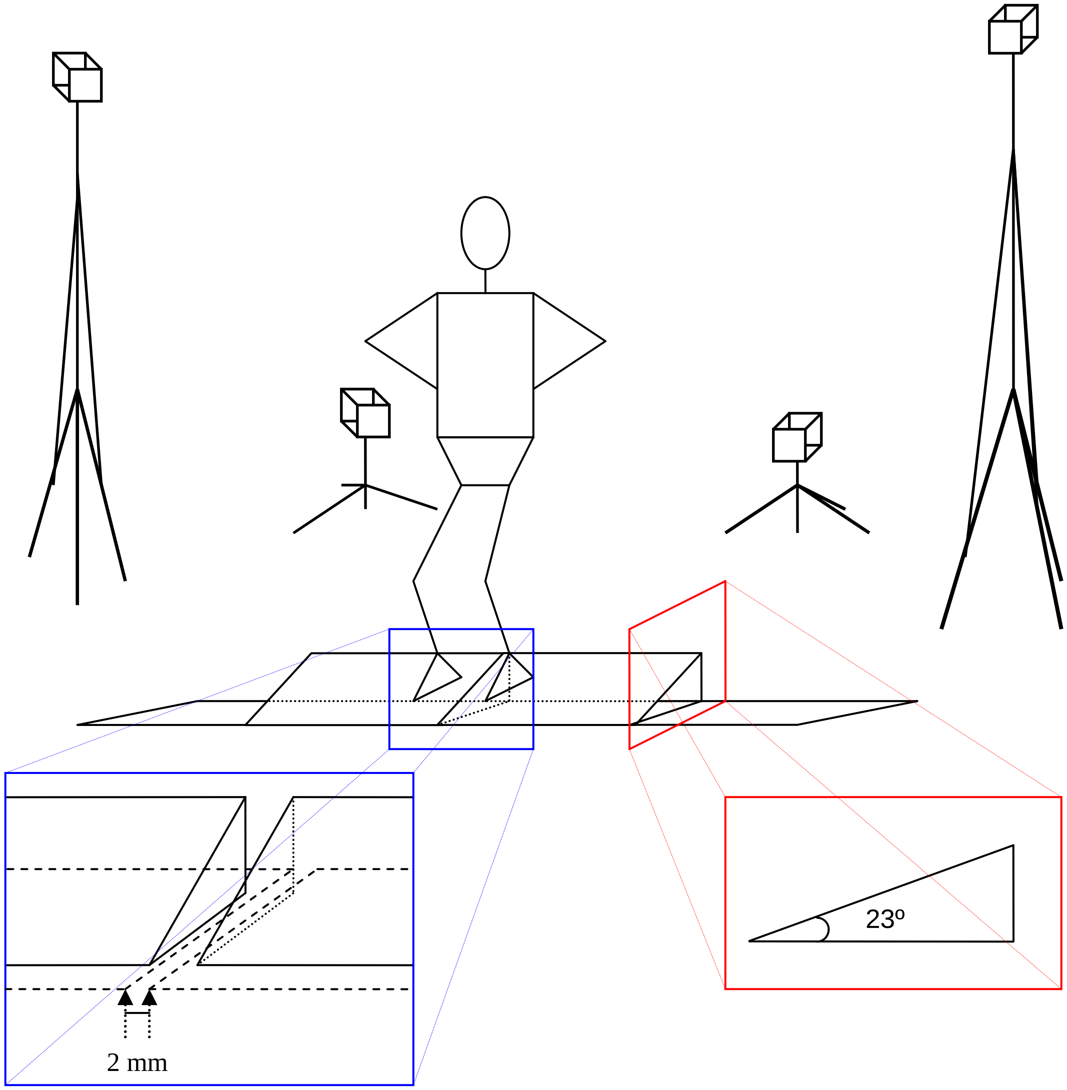


Figure 3.3.2 Schematic representation of the experimental set up during hopping. Two force plates, decline boards and a representative four cameras aimed at the capture volume. Expanded image of the two decline boards over the force plates (dashed blue lines) shows the 2 mm gap between the force plates (distance between two arrows) and gap between the decline boards. Expansion of the single decline board (dashed red lines) shows the angle to be 23º.

3.3.3 Protocol

3.3.3.i Participant Preparation

Each participant was prepared for testing by adhering light reflective markers (14mm diameter) using double-sided tape as per the VICON plug-in-gait model (VICON Motion Systems, Oxford, UK) (Appendix E).

The skin was debrided using gauze (10 x 10cm) (Smith & Nephew) and skin preparation gel (Nuprep) before adhering surface EMG electrodes. Single differential bipolar Ag-AgCl surface electrodes (Noraxon, USA) that were 4mm in diameter with a centre to centre distance of 20mm, were adhered parallel to the muscle bellies of the following lower limb muscles on left and right sides: gluteus medius (GM), vastus lateralis (VL), biceps femoris (BF), medial gastrocnemius (MG), soleus and tibialis anterior (TA) (Figure 3.3.3). Electrode placement was confirmed by asking the participant to perform and isometric contraction while standing. The muscle belly was identified by palpation and electrodes adhered parallel to the line of the muscle fibres.

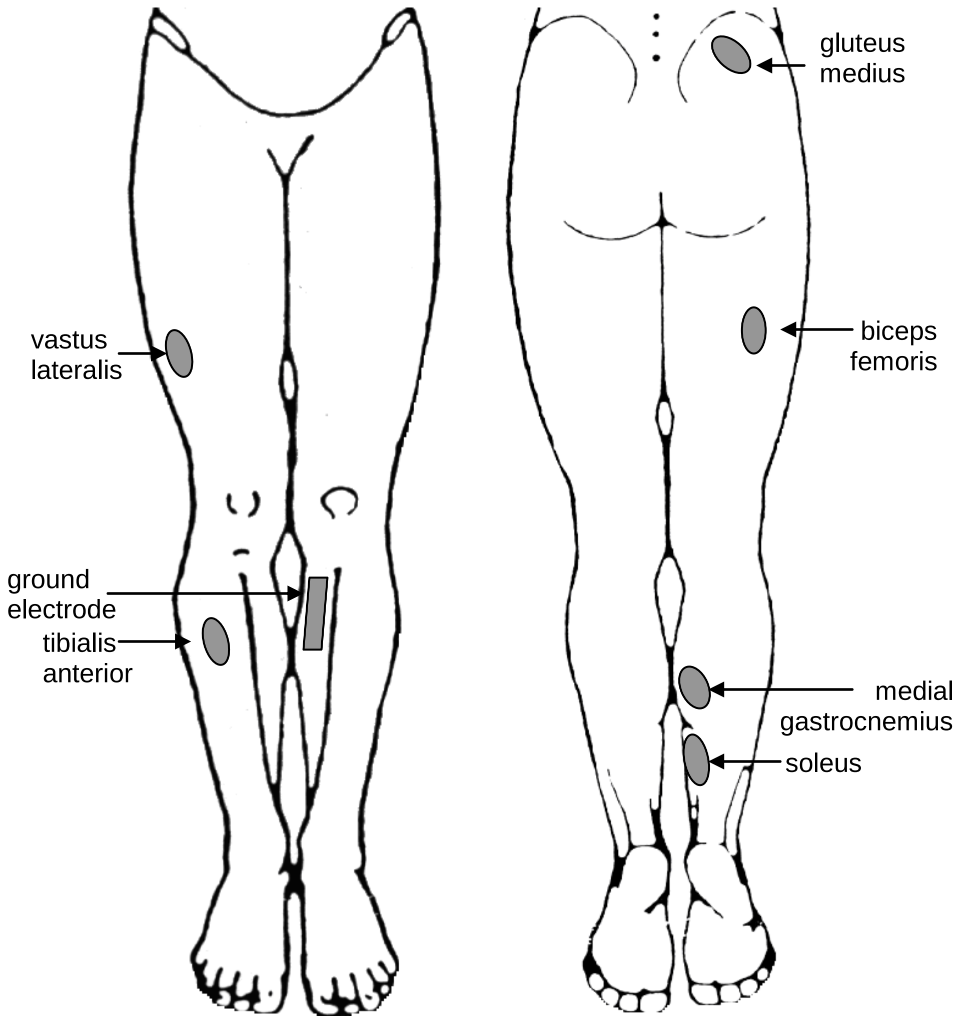


Figure 3.3.3 Schematic representation of surface electromyography electrodes adhered over muscle bellies of the right leg and a ground electrode adhered over the left tibia that is relatively superficial. Each shaded ellipsoid represents a bipolar electrode pair.

The choice of electrodes, process of skin preparation, positioning of the patient (standing) during electrode placement, sensor location (Figure 3.3.3) and the fixation of electrodes were maintained with reference to the sEMG for a non-invasive assessment of muscles guidelines (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000).

3.3.3.ii Motor Tasks

Two hopping tasks were included; DL hopping on-the-spot and AL hopping on-the-spot. Double-leg hopping was defined as hopping with both legs being in phase for contact and flight periods.

Both tasks (DL and AL) were performed with the participant's hands on their waist to prevent upper limb motion and thereby prevent upper limb artifact. This is consistent with the majority of research on sub-maximal hopping literature with the strategy to avoid upper limb motion during hopping (Dyhre-Poulsen, et al., 1991; Hobara, et al., 2008; Rapoport, Mizrahi, Kimmel, Verbitsky, & Isakov, 2003). Participants were instructed to hop by using a forefoot first strategy as a heel first strategy is shown to modify joint contributions demonstrated in drop jump landings (Kovács, Tihanyi, Devita, Rácz, Barrier, & Hortobágyi, 1999).

Each hopping task (DL and AL) was performed on a flat surface and a decline (23º) surface. Each hopping task (DL and AL) was performed at a self-selected hopping pace and at a forced hopping pace of 1.8 Hz. The forced pace of hopping at 1.8Hz was chosen to control for any variation in hopping frequency between hopping conditions for the factors of surface and task. Furthermore, 1.8Hz was considered a sub-maximal hopping frequency and approximated the common hopping frequency of athletes (2.0Hz) (Dyhre-Poulsen, et al., 1991) and non-athletes (Farley, et al., 1991; Funase, et al., 2001; Hobara, et al., 2007). The DL and AL hopping tasks on the flat and decline surfaces at a self-selected hopping pace were always performed prior to the tasks being performed at forced pace. This was to ensure that a true self-selected pace of hopping frequency was selected, thus minimising any influence of the forced hopping pace. An alteration in the self-selected hopping frequency has been shown if a prior controlled hopping frequency is performed first (Padua, et al., 2006). The order of hopping task (DL or AL) and hopping surface (flat or decline) allowed for a combination of four conditions (DL hopping on the flat, DL hopping on the decline, AL hopping on the flat and AL hopping on the decline) and the order of performance was block randomised to reduce the risk of systematic error. This was achieved by allowing the participant to select a numbered card at random that referred to one of the four conditions. Thus, each participant performed hopping under a total of 8 separate conditions based on hopping task (DL or AL), hopping surface (flat or decline) and hopping pace (self-selected or forced pace) in a single testing session on the same day.

Forced pace hopping was performed by the use of an audible tone (0.25s duration at 108 beats/minute) from a metronome (custom built software using LabView 7.0, National Instruments, US) that was played directly through the speakers of the laptop computer (Dell Precision M20, Malaysia). The participant was instructed to land on the sound of the tone thereby controlling the hopping pace to 1.8 Hz.

All hopping tasks were performed on dual, adjacent force plates to allow recording of the force plate signal from each leg in isolation (Figure 3.3.2). The use of dual force plates has previously been used in assessing propulsion and landing phases during a simulated block jump task in volleyball players (Hughes & Watkins, 2008) and during hopping on a flat surface (M. Voight, et al., 1995a).

3.3.4 Data Collection

Prior to data collection a warm-up and familiarisation period was performed. The participant was instructed to hop on-the-spot at a comfortable pace for up to 30 seconds. Then, a static trial was captured with the participant standing still, away from the force plates. Following instruction and demonstration from the investigator, each hopping condition was practiced between two to four times. The participant had to maintain his hands around the iliac crests, to hop on-the-spot with each leg on a separate force plate and to avoid movement forwards, backwards, or laterally.

For trials that were recorded, the participant was instructed to hop for between 15 to 30 seconds for each trial. The participant was verbally instructed to start and stop the task. Between 1 and up to 3 trials were recorded and stored (Nexus 1.3) for each hopping condition. More than one trial was recorded for a task if the participant was observed to move off the force plates forwards, backwards or laterally onto a single force plate. The force plates, motion capture system and EMG signal were calibrated prior to data collection of

each trial by instructing the participant to remain off the force plates and remain completely still and the calibrate function entered on the Nexus (version 1.3) interface.
Nexus software (version 1.3) recorded motion capture, analogue signals from the two force plates and analogue signals from the EMG system synchronously. Data were stored on a personal desktop computer (Hewlett-Packard XW4300 Work Station).

3.3.5 Data Processing

Force plate, EMG and kinematic (joint angles) data were confirmed for a single trial for each task performed, by visual inspection of raw data. Static (participant maintained a static posture in the capture volume) and dynamic trials were labeled by their corresponding anatomical landmark (Appendix E). For dynamic trials, gaps in marker trajectories were manually filled for each trial using either a spline or pattern fill function on an updated version of Nexus (Nexus 1.4) from the version on which data was collected (Nexus 1.3). Maximum fill was no more than 10 frames.

Kinematic and kinetic data were filtered using a 2nd order zero-lag Butterworth filter at 12 Hz and 80 Hz respectively. Finally, static and dynamic models were run for the captured static and dynamic trials respectively. The static model was always run before dynamic trials to define joint centres (Nexus version 1.4). Data including the time, EMG signals, ground reaction forces (x, y and z axes) from each force plate, joint kinematics and joint moments were exported to excel spreadsheets (Microsoft Office Excel 2003).

3.3.5.i Inverse Solution

Nexus software (version 1.4) performed the inverse solution to produce values for joint moments prior to exporting the data into an excel spreadsheet. The inverse solution to calculating joint moments is briefly described in Appendix F (Winter, 1990).

3.3.5.ii Hopping on a Decline Surface – Correction for Joint Moment

Calculation of joint moment is reliant on an accurate measure of the centre of pressure on each force plate during the contact phase. During decline surface hopping, the participant is visualised as being above the force plates within the Nexus software (version 1.4) as the global origin is aligned with reference to the force plates that are level with the floor. Thus, the participant is seen to hop above the force plates even though he is contacting the decline boards. The centre of pressure is therefore not accurately represented with respect to the point of contact of the participant on the decline board. This influences the calculation of joint moments using the inverse solution and may cause a systematic bias to the derived data set from hopping on a decline board. Thus, a pilot study was performed to compare two methods of height correction of force plates for calculation of joint moments when hopping on a decline surface to reduce the time of data processing. This method is detailed in Appendix G.

3.3.6 Derived Variables

All derived variables (Table 3.3.1) were calculated for the left and right legs separately as gait events, determined by the vGRF and were measured for each leg independently.

Table 3.3.1 Derived variables including kinetics, kinematics and muscle activation amplitude characteristics.

Derived Variable	Unit
Hopping Frequency	Hertz (Hz)
Vertical height displacement of centre of mass	Metres (m)
Peak vertical ground reaction force	Newtons (N)
Vertical leg stiffness	$N.mm^{-1}$
Impulse	$\Delta F.t$
Joint excursion (ankle, knee and hip) during braking	Degrees (º)
Normalised joint moment^{BM} (ankle, knee and hip) during braking	$N.mm.kg^{-1}$
Normalised joint stiffness^{BM} (ankle, knee and hip) during braking	$N.mm.kg^{-1}.radian^{-1}$
Joint angular velocity (ankle, knee and hip) during braking	º/s
Normalised joint work^{BM} (ankle, knee and hip) during braking	$J.kg^{-1}$
Peak activation amplitude	Volts (V)
Normalised pre-loading activation amplitude	%

BM – Normalised to body mass (kg)

3.3.6.i Temporal Gait Events

Each single trial contained repeated hopping of between 30 and 50 hops. A single hop was defined as a flight phase and the subsequent contact phase. Flight phase (T_f) was from toe-off (TO_1) to initial contact (IC) and contact phase was from IC to toe-off (TO_2) (Figure 3.3.4) consistent with previous studies (Blickhan, 1989; Farley, et al., 1999; Rousanoglou & Boudolos, 2006). From the point of IC to peak vertical ground reaction force (vGRF) was defined as the braking phase. The gait events of interest for each cycle were: IC and peak vGRF for each hop on each leg in all DL and AL hopping conditions.

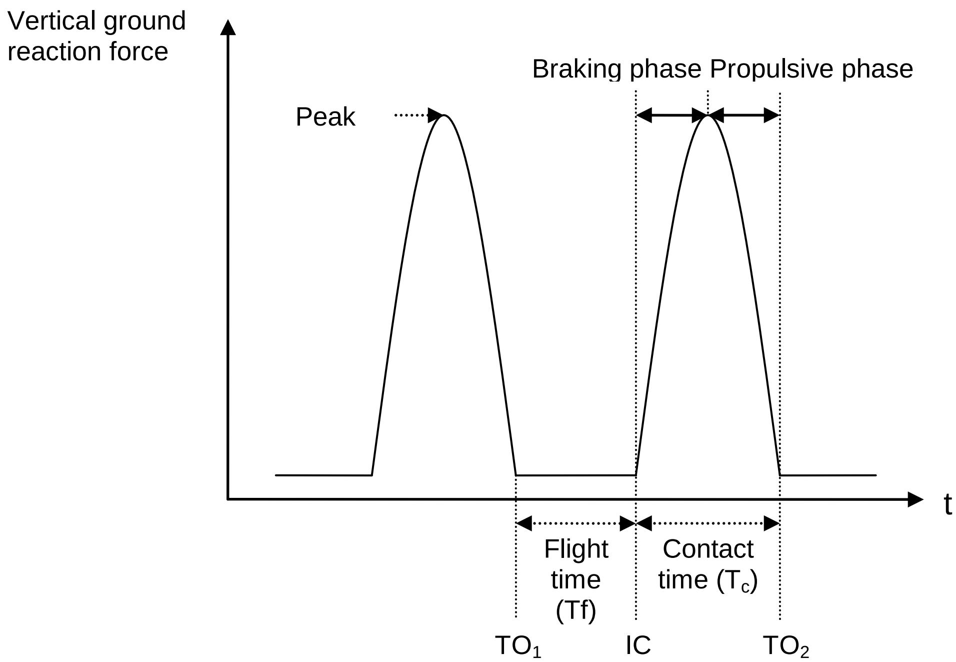


Figure 3.3.4 Schematic representation of a force signal in the z axis (vertical ground reaction force) collected from a single force plate with respect to time. Flight phase (T_f) is from toe-off$_1$ (TO_1) to initial contact (IC)). Contact phase (T_c) is from IC to TO_2 and divided into braking phase (IC to peak F_z) and propulsive phase (peak F_z to TO_2).

The flight phase refers to the period of time that the COM is not in contact with the ground. For DL hopping the left and right sides are observed to be in flight and contact synchronously (Figure 3.3.5). Gait events (IC, peak vGRF and toe-off) were determined for the left and right legs independently.

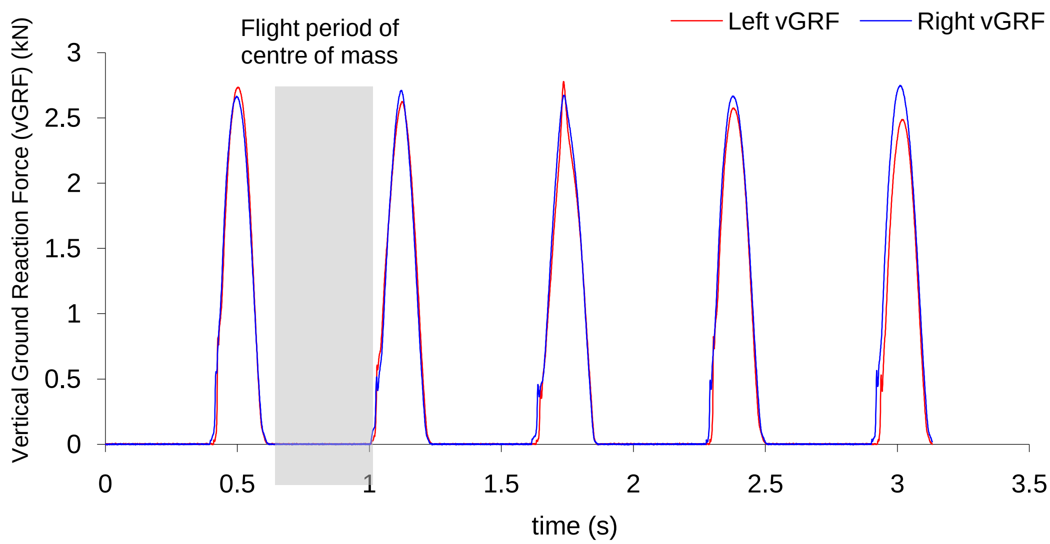


Figure 3.3.5 Example from the current study of a recording of the vertical ground reaction force for 5 consecutive hops on left and right legs, taken from a single trial performed by a participant during double-leg hopping on a flat surface at a self-selected pace. Toe-off and initial contact occur synchronously for left and right legs.

For AL hopping the COM is in flight only briefly while one foot has completed toe-off and the contralateral foot prepares to contact the ground (Figure 3.3.6). The flight period was the period that there was no contact with either left or right legs, preceding a single leg contacting the ground. The gait events were obtained for the left and right legs, independently and with respect to the COM, i.e.; when there was no contact between the ground and the individual with either leg.

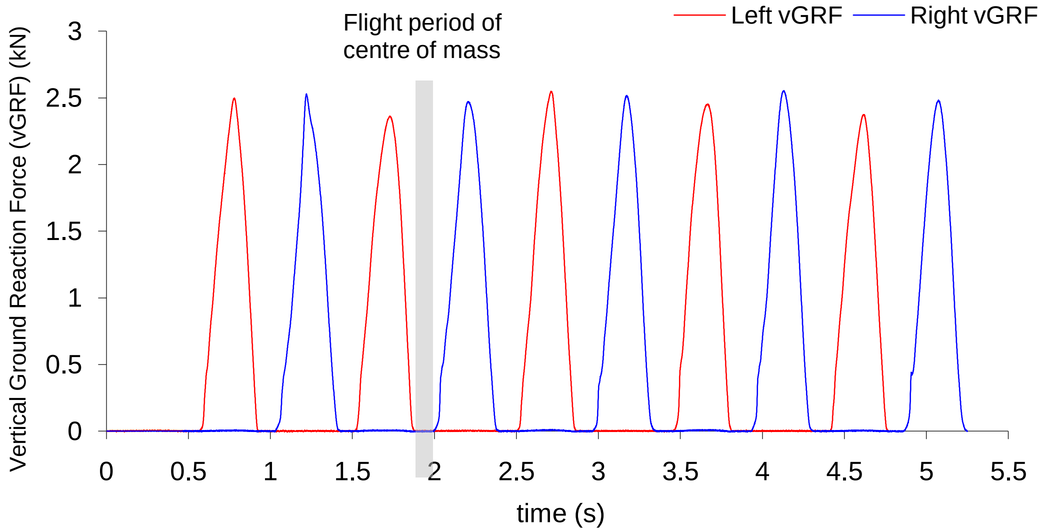


Figure 3.3.6 Example from the current study of a recording of the vertical ground reaction force for consecutive hops on left and right legs during alternate-leg hopping on the flat surface at a self-selected pace. Time between the toe-off of the left leg and initial contact of the right leg is the period of flight for the centre of mass prior to right foot contact (grey shaded area). This is the flight phase of the participant when there is no contact with the ground.

To determine gait events, the vGRF at IC, peak vGRF and vGRF at toe-off were determined by manual inspection of the vGRF-time graph based on the premise that there will be an increase in vGRF from '0' as the foot strikes the force plate (Figure 3.3.7). The use of GRF in determining gait events is considered the gold standard (A. H. Hansen, Childress, & Meier, 2002; Hreljac & Stergiou, 2002; Zeni Jr., Richards, & Higgninson, 2008). The protocol used to determine gait events was to identify the point at which there was a change in vGRF. The linear region represented the time in flight when there was no change in vGRF values. The vGRF value at IC was identified as the first value before there was a continuous increase in vGRF to the peak vGRF value. Peak vGRF was identified by determining the maximum value for each hop during the contact phase. The vGRF at toe-off was chosen as the lowest value before there was a change from the linear region of the force-time curve, representing flight phase, for that leg (Figure 3.3.7a-d).

Determining vGRF at toe-off, IC and peak vGRF then provided the gait events for each hop. From these events temporal data for each hop including the flight, braking and propulsive periods were determined. This allowed for processing of derived variables (Table 3.3.7).

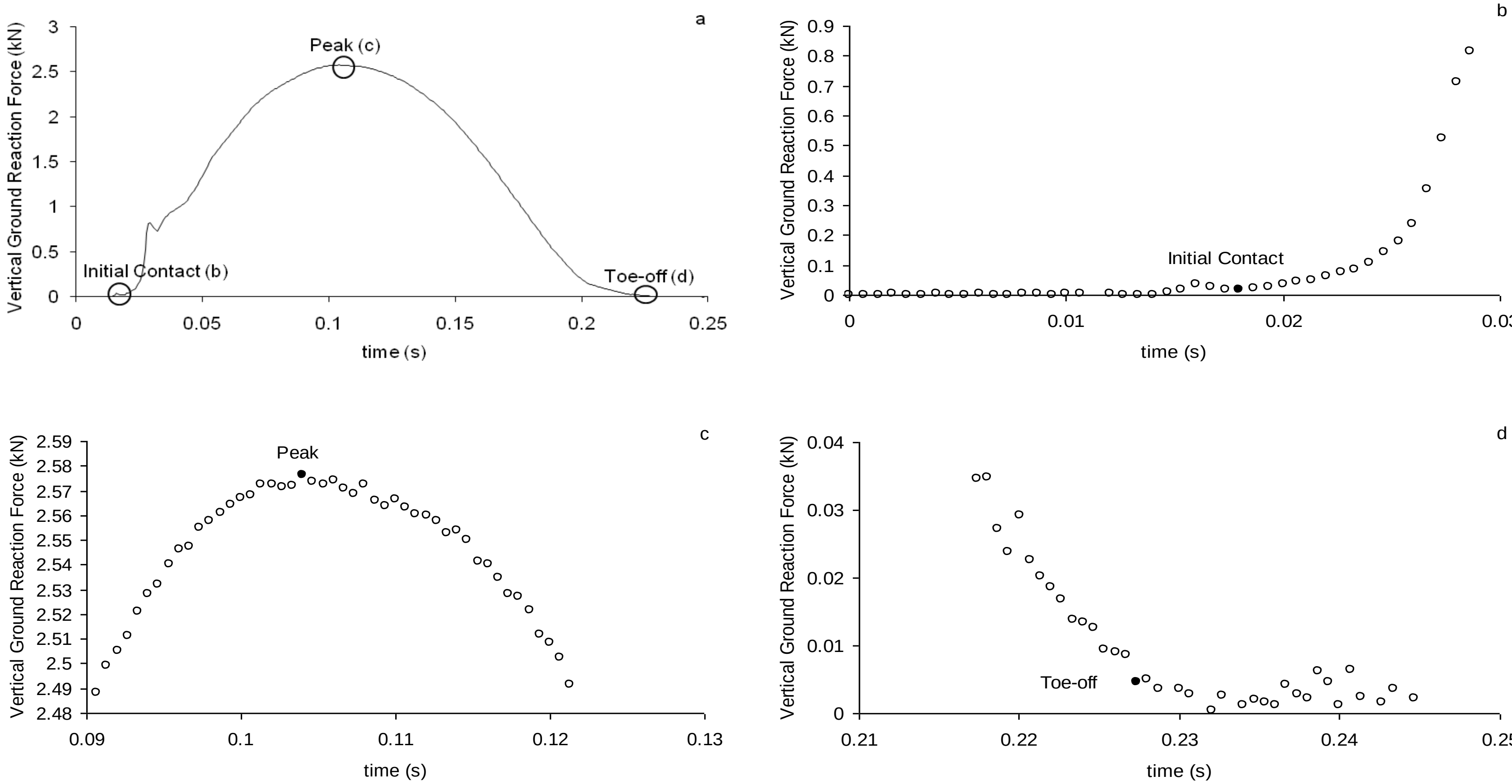


Figure 3.3.7 Force-time curve (a) for a single hop. Enlargements of the vertical ground reaction force (vGRF) value at initial contact (b), peak (c) and toe-off (c) were determined by manual inspection of the vGRF-time recording. vGRF was a sampled at 1500Hz.

3.3.6.ii Hopping Frequency

Hopping frequency was calculated for each trial by truncating the vGRF-time data and included only hops where there were complete flight and contact phases from the first to last hop, regardless of the total number of hops recorded (Figure 3.3.8). Thus, the number of complete hops was counted and divided by the time over which they were performed, yielding the hopping frequency in Hertz. During DL hopping tasks the left and right legs act in parallel by being in flight and contact phases at approximately the same time. In AL tasks the left and right legs alternate such that while one leg is in flight the contralateral leg is in contact. It is important to note that during AL tasks the COM is contacting the ground with one leg half the number of times and with the contralateral leg the other half number of times over the measured time. Thus, the hopping frequency of the COM is the sum of hops on the left and right over a measured period of time, while during DL hopping the hopping frequency is equal to the hopping frequency of the left or right leg only.

Hopping frequency was calculated for the left and right legs during DL and AL hopping tasks to allow for inter-side comparison. To allow comparison across hopping conditions, the hopping frequency of the COM was calculated for DL and AL tasks.

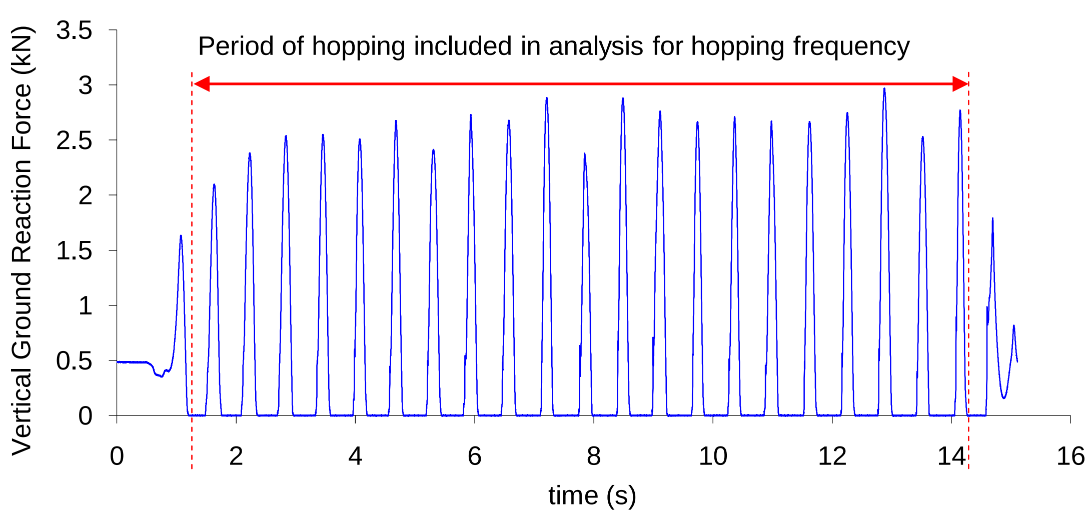


Figure 3.3.8 A recording of the vertical ground reaction force trace collected for the right leg during double-leg hopping on a flat surface at a self-selected pace. Data is truncated (dashed vertical red lines) to include only complete cycles (flight and contact phases) of hopping with truncation of data at the start and end of the recording.

Calculation of vertical leg stiffness requires values of vertical displacement and peak vGRF. To generate values for vertical height displacement and peak vGRF for each leg during each trial, ten consecutive hops were chosen for DL and AL hopping. Choosing consecutive hops allowed for an inclusion of natural variation in data. This was seen as more representative of hopping under differing conditions of task, surface and frequency rather than either choosing a single hop or choosing the 'best' hops based on one or more factors from the data reducing the risk of selection bias.

Prior to the calculation for derived variables, a selection of consecutive hops was made by the primary investigator by visual inspection of the vGRF-time curves. Five consecutive hops were selected for calculation of EMG derived variables and measures of vertical stiffness and ten consecutive hops were selected for kinematic and kinetic derived variables consistent with previous studies (Granata, Padua, & Wilson, 2002; Hobara, et al., 2010bHobara, 2008 #1521; Padua, et al., 2005).

The consecutive hops chosen for each trial were based on the two factors of having approximately equal peak vertical ground reaction forces between hops and approximately equal peak vertical ground reaction forces between left and right sides. Care was taken to ensure that each hop had a smooth increase in vGRF at IC. If a hop was discarded from inclusion then consecutive hops (five or ten) before or after the hop that was discarded were inspected for equality in amplitude of vGRF and a smooth increase in vGRF at IC.

Once the consecutive hops were chosen the gait events were identified and time periods for flight phase, braking phase, propulsive phase and contact phase identified for each hop. The peak vGRF was also calculated by subtracting the vGRF at IC from that at peak vGRF for each hop. This yielded the 'change' in GRF during braking phase as GRF at IC was close to, but not exactly zero. The mean values for flight, contact, braking, propulsive phases and peak vGRF were calculated across the 10 consecutive hops.

3.3.6.iii Kinetic Variables

3.3.6.iii - Vertical Height Displacement

Vertical height displacement during braking phase was calculated for each of the five selected consecutive hops on the left and right sides. There were no missing data points for vGRF that were used for calculation of vertical height displacement.

The trajectory of motion of the spring-mass model assumes a height change that follows a pattern described by simple harmonic motion in that it is sinusoidal in trajectory over flight and contact phases (Figure 3.3.9). When there is a change in direction of trajectory from positive to negative (z axis) of the COM, it may be assumed that velocity (v) is 0m/s at some point between positive and negative height displacements as the mass has to decelerate to 0m/s before accelerating again. The spring-mass model has been shown to model the mechanical behaviour accurately during hopping (Blickhan, 1989; Brughelli & Cronin, 2008). The parabolic trajectory of the vertical height displacement curve is true for flight phase where the individual is subject only to gravitational force and resistance due to air. Air resistance is assumed to be zero for the purposes of this investigation as the participant is in flight for very short periods and thus does not reach high velocities prior to impact.

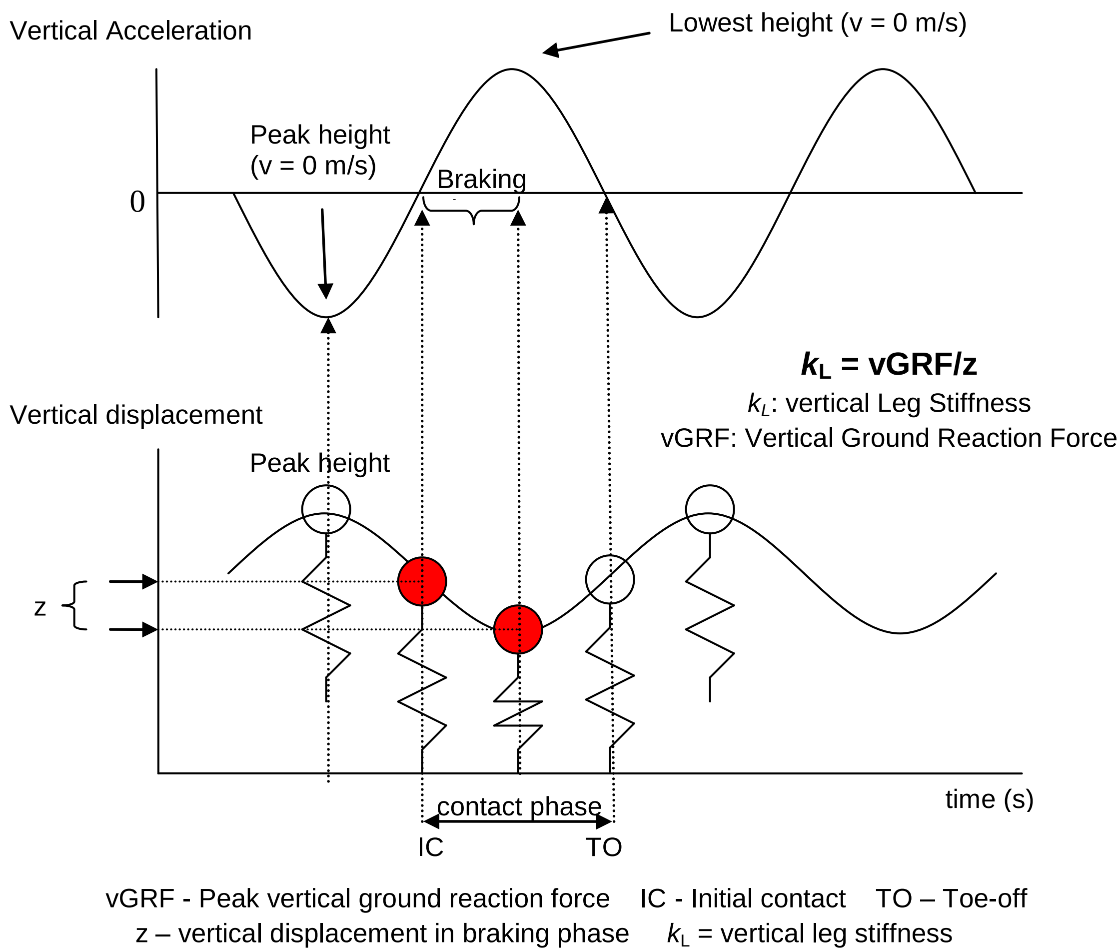


Figure 3.3.9 Schematic representation of the sinusoidal trajectory of the mass being displaced along the vertical axis during hopping where v = 0 m/s when there is a change in direction of displacement. Vertical leg stiffness (k_L) is the change in force per unit displacement divided by the change in height of the centre of mass during braking phase (red shaded mass).

In this investigation by measuring vGRF and being able to identify gait events accurately, the assumption did not have to be made that the vertical displacement-time function is sinusoidal in the contact phase. As vGRF was measured directly and the time during braking phase was able to be determined accurately, vertical height displacement during braking phase was able to be determined following the principles of Newtonian mechanics of falling masses and has been used in previous studies (Flanagan & Comyns, 2008; Lloyd, et al., 2009).

Thus, from first principles, the displacement of the COM from IC to peak vGRF was estimated by the action of gravity acting on the mass during the second half of flight time (maximum height, where velocity is zero, to IC), in that flight time is known and vertical height displacement in flight to time curve was parabolic.

$$z_f = \frac{1}{2} \cdot g \cdot \left(\frac{T_f}{2}\right)^2 \text{ ....equation 1}$$

Where,

z_f was the vertical displacement of a mass from peak height to IC (second half of flight phase), g was acceleration due to gravity (9.81m/s^2) and T_f was the total time in flight.

The velocity at the point of IC is estimated by the acceleration due to gravity

$$v_i = \sqrt{2.g.z_f} \text{ ....equation 2}$$

Where,

v_i is the velocity at IC, g is acceleration due to gravity (9.81m/s^2) and z_f is the vertical displacement of the mass from the peak height to IC.

Then, the displacement of the COM from IC to peak depression (braking phase) is determined by multiplying the average velocity during braking phase to the time from initial contact to peak vGRF

$$z_c = \frac{1}{2}\left(v_i + v_f\right) \cdot \left(\frac{T_c}{2}\right) \text{......equation 3}$$

Where,

z_c is the vertical displacement of the mass from IC to peak F_z (braking phase), v_f is the final velocity at peak vGRF (0 m/s) and T_c is time in contact.

3.3.6.iv - Vertical Leg Stiffness

Hooke's Law states that a deformable body may store and return energy based on the formula *F = kx* where the force (*F*) required to deform a body is proportional to the proportionality constant (*k*) and the distance (*x*) that the object is deformed on the condition that the body is not permanently deformed (R. J. Butler, Crowell III, & Davis, 2003). A number of methods have been used for calculation of vertical stiffness (R. J. Butler, et al., 2003) and this study simply describes the ratio of change in vGRF to the change in displacement of the COM during braking phase, where motion is assumed to occur only along the z axis in a linear manner (T. A. McMahon & Cheng, 1990). In this case the measure of vertical stiffness relates to linear systems such as springs and was noted as being a measure of ‘quasi-stiffness’ as it does not model each element that contributes to stiffness including ligament, tendon, muscle, bone and cartilage as well as neural contributions in controlling a multi-joint system (M. L. Latash & Zatsiorsky, 1993).

Vertical stiffness has been simply modeled as deformation of the COM during the braking phase and is termed vertical leg stiffness (Kuitunen, Komi, & Kyroläinen, 2002; T. A. McMahon, et al., 1990). It is understood that the leg is considered the spring element and that deformation will primarily occur as a shortening of the leg. Hence, previous studies have estimated leg spring deformation and calculated ‘leg stiffness’ (Kerdok, Biewener, McMahon, Weyand, & Herr, 2002; P. J. McMahon, et al., 1987; T. A. McMahon, et al., 1990; Morin, G., Kyröläinen, Jeannin, & Belli, 2005). Surface stiffness has been shown to affect leg stiffness (Farley, et al., 1998; Ferris, et al., 1997; Ferris, Louie, & Farley, 1998), however, in this study with a repeated measures design and no change in surface stiffness between hopping conditions, the measure of vertical leg stiffness was seen as sensitive to changes due to hopping condition only. There has been also shown to be a strong correlation between vertical stiffness and leg stiffness (Morin, et al., 2005).

In the current study, the method used to calculate vertical leg stiffness was the reference method and described below. A field test method utilising only flight and contact time data for hops has been used previously to calculate vertical leg stiffness (Dalleau, Belli, Viale, Lacour, & Bourdin, 2004). Values obtained for force, vertical height displacement and vertical leg stiffness were compared between the reference method and field test method (chapter 4).

Vertical leg stiffness (k_L) during braking phase was determined by the ratio of force to vertical height displacement in braking phase (z_c) (T. A. McMahon, et al., 1990) given by the following equation:

$$k_L = \Delta \text{ vGRF}/z_c \text{.........equation 4}$$

Hence, k_L was calculated by knowing the flight time, braking time and peak vGRF.

3.3.6.iv Joint Variables

3.3.6.iv - Joint Excursion, Stiffness and Work

A selection of 10 hops was equal or greater than the number of hops analysed in previous studies (Farley, et al., 1998; Farley, et al., 1999; Hobara, et al., 2008; Padua, et al., 2006; Padua, et al., 2005) and deemed representative of hopping under each condition.

The reason to increase the selected number of hops for analysis was based on missing kinematic data for some hops and therefore missing data for values of joint moment (Figure 3.3.10) that are calculated from centre of pressure and GRF data. Missing kinematic data were at the same periods for all joints in each trial. Any hop where there was no clear identification of a joint angle at IC or peak vGRF was discarded. A hop either before or after the 10 chosen hops was then included for analysis to maintain a sample of 10 hops. For some trials there were less than 10 hops. Thus, for these trials all hops that allowed identification of a joint angle at IC and at peak vGRF were included.

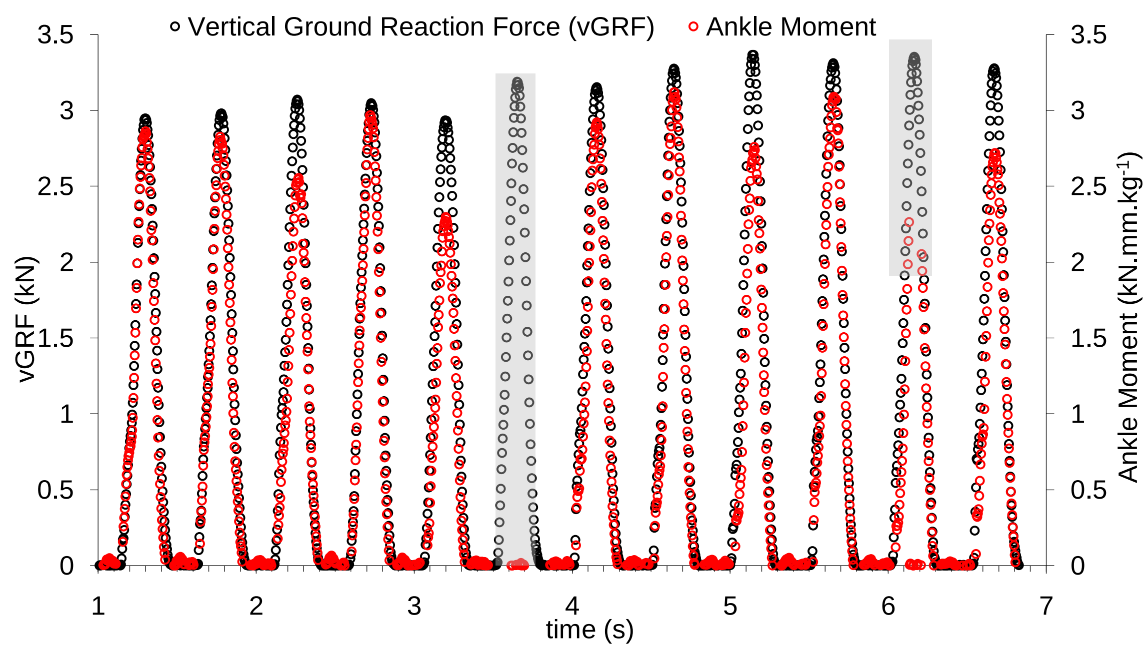


Figure 3.3.10 An example from the current study of a recording for a single participant of the total vertical ground reaction force (black circles) and left ankle moment (red circles) during double–leg hopping on a flat surface at a self-selected pace showing regions of missing data points (shaded grey areas) during the contact phase of the left leg. These hops were deleted and the next complete hop with no missing data points was selected for inclusion for each trial until there were a total possible 10 hops selected.

There were a total possible of 160 hops for analysis for each participant (i.e.; 20 hops for a single trial (10 left and right each) and 8 hopping trials). A total of nine participants in this study yielded a total possible 1440 hops that could have been included for analysis. Only two trials for two participants included less than 10 hops with a total of 3 hops not included for analysis. There were 71 hops that had to be replaced from the total of 1437 that were selected. Thus, only 0.2% of hops were missing and 4.9% of hops were replaced due to missing data points for joint moments. Joint moment data was missing due to either missing data point for GRF or joint kinematics. As the number of hops with missing data was small (4.9%) and able to be substituted using data for successive hops, this was not considered a significant confounding variable.

Once the hops were selected, the gait events were identified for each hop and the following variables determined for each hop during braking phase, calculated as the difference in values between initial contact to peak vGRF: time, joint excursion and joint moment.

In this study the term joint stiffness refers to the normalised joint moment ($Nm.kg^{-1}$) (normalised to body mass (kg)) divided by the joint excursion (radians) to yield joint stiffness ($Nm.kg^{-1}.rad^{-1}$) consistent with previous studies (Farley, et al., 1998; Farley, et al., 1999; Günther & Blickham, 2002; Hobara, et al., 2010b; Hughes, et al., 2008; M. L. Latash, et al., 1993; Müller, et al., 2010). Joint moment and mechanical energy were normalised to body mass to allow a pooling of data across all participants who were of different stature.

Joint angular velocity (degrees/s) was calculated by dividing the joint excursion (degrees) by the braking phase time (s) (Yeow, Lee, & Goh, 2010).

Joint work ($J.kg^{-1}$) during braking phase at the ankle, knee and hip joints was calculated using the following steps as defined previously by Bisseling et al. (2007) and Yeow et al. (2010) (Bisseling, et al., 2007; Yeow, et al., 2010).

1. Power (Watts or $Joules.kg^{-1}.s^{-1}$) was determined by multiplying joint moment ($Nm.kg^{-1}$) and joint angular velocity (radians/s) (converted to rad/s from degree/sec) such that each value represented the mean during the braking phase of each hop. Note that moment was

normalised to body mass (kg). All derivates of moments were therefore also normalised to body mass.

2. The product of power ($J.kg^{-1}.s^{-1}$) and time in braking phase (s) yielded the mechanical work done ($J.kg^{-1}$) (where energy (J) is normalised to body mass (kg)) in braking phase. It is common to denote the eccentric phase that is the braking phase as being negative work, as mechanical energy is absorbed. For convenience, as the braking phase was the only period of interest in this investigation, mechanical work done has been expressed as a positive value although it is understood that the value represents the magnitude of mechanical energy absorbed rather than produced in braking phase. Mechanical energy is normalised to body mass in kg as moments were normalised to body mass to allow pooling of values for all participants and comparison across hopping conditions.

The mean of values for each of the derived variables (Table 3.3.1) was calculated across the total number of selected hops.

3.3.6.v Electromyography Variables

The five consecutive hops chosen for calculation of vertical leg stiffness were also used for analysis of EMG derived variables for each selected hopping trial. The selection of five hops was consistent with previous studies evaluating EMG characteristics during hopping under different conditions (Dyhre-Poulsen, et al., 1991; Funase, et al., 2001; Hobara, et al., 2008; Melvill-Jones, et al., 1971).The EMG signals from 6 muscles on each side over 8 hopping conditions were recorded for each participant yielding a total of 96 possible recordings. For 9 participants the total possible number of EMG recordings was 864. Five recordings were not able to be used due to a loss of signal during hopping, loss of recording during hopping or signal recording that appeared to have excessive noise. Thus, only 0.6% of EMG recordings were unable to be used in data analysis of EMG signals.

Electromyography data were collected parallel to the GRF data using Nexus 1.3 at the same sampling rate of 1500 Hz as EMG (analogue) signal. Electromyography data was truncated at the same time intervals as vGRF data that defined the first and last hops included in the analysis and stored in a separate text format file. Each trial was analysed using custom built processing software for filtering and processing (Labview 8.0, National Instruments). The

EMG signal was band-pass filtered (20-500 Hz) using a 4th order zero-lag Butterworth filter and full wave rectified. Linear envelopes were then created for EMG and vGRF data using a 2nd order zero-lag Butterworth low pass filter (cutoff frequency 200 Hz). Gait events of interest were IC and peak vGRF.

3.3.6.v - Peak Activation Amplitude

The maximum EMG value was identified in a 40ms moving window during braking phase (from IC to peak vGRF) and defined as the peak activation amplitude of the muscle during braking phase (Figure 3.3.11). The average of the peak values over the five consecutive hops during braking phase was calculated for each muscle. The average rectified value over a 40ms epoch in this study (Figure 3.3.11) is similar to that used in other studies on hopping (Hobara, et al., 2010a; Padua, et al., 2006) and the use of a root mean square over a 10ms sliding window in drop landing tasks (Blackburn & Padua, 2009). Other studies have used the average EMG signal over the complete stance phase in hopping (Moritz, et al., 2005). In the current study peak amplitude was defined as the peak value calculated over the sum of values over a 40ms window occurring during braking phase. This was deemed more sensitive to changes in the braking phase only and less likely to detect EMG signal due to pre-loading activation, activation in the concentric phase of muscle action or phase shift that may occur as movement was rapid in hopping. Therefore, the use of a 40ms compared to a 10ms moving window was not only consistent with previous studies it was also deemed more representative of the EMG signal during braking phase. Statistical analysis was performed on the raw peak activation amplitude values for each trial across hopping conditions for each muscle on the left and right sides with a within-subject design (3-factor repeated measures ANOVA).

The repeated measure design of this ensured that difference in peak activation amplitude was due to a difference across hopping conditions. Maintaining electrode placement in the single testing session ensured that the peak activation amplitude was not affected due to differences in electrode placement. Short duration hopping tasks and adequate rest periods between trials ensured that fatigue did not affect the peak activation amplitude.

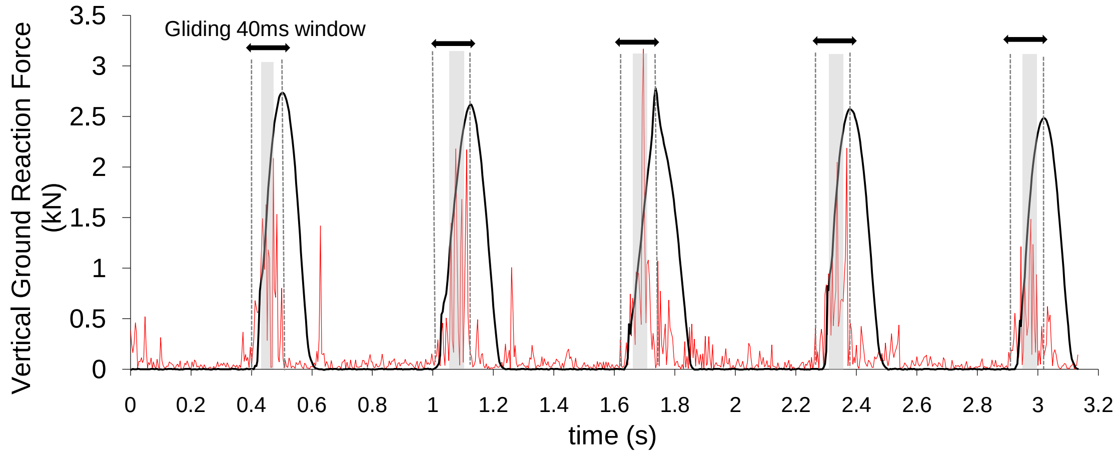


Figure 3.3.11 An example from the current study of the left vastus lateralis surface electromyography signal (full-wave rectified and band-pass filtered 20-500Hz) (red) and vertical ground reaction force (vGRF) signal recording (black) of the left leg, over five consecutive hops during double-leg hopping on a flat surface at a self-selected pace. Peak activation amplitude was defined as the peak value calculated over the sum of values over a moving 40ms window during braking phase (initial contact to peak vGRF) (grey shaded area and region between dashed vertical lines).

The coefficient of variation (CV) may be reduced when amplitude normalised values are calculated and may remove variance that is considered normal in the biological system (Allison, Marshall, & Singer, 1993). It was also demonstrated that normalising EMG signal to sub-maximal eccentric contractions of the QF did not provide any benefit and reduced the reliability of the measuring the root mean square EMG signal (Finucane, Mayhew, & Rothstein, 1994). Thus, for the purposes of this investigation the ‘raw’ EMG signal was used as it allowed comparison of the signal sensitive to high speed movement (hopping) and specific to the eccentric (braking) phase only. The consistency of variance of measures of peak activation amplitude were ensured between hopping conditions prior to statistical analysis and the within-subject design ensured that any significant changes were as a result of a change due to hopping condition only.

3.3.6.v - Pre-Loading Activation Amplitude

The EMG data for the MG muscle only, was averaged over the 100ms preceding IC to provide a value of pre-loading activation amplitude (Figure 3.3.12) consistent with previous studies (Bonnard, Sirin, Oddsson, & Thorstensson, 1994; Moritani, Oddsson, &

Thorstensson, 1990). An ensemble average of the values over the five consecutive hops was calculated for pre-loading activation amplitude of MG on left and right sides.

Pre-loading activation of the MG muscle was of interest in this study as it has been shown to be sensitive to changes across hopping conditions (Funase, et al., 2001; Hobara, et al., 2007; Melvill-Jones, et al., 1971) and running (Müller, et al., 2010).

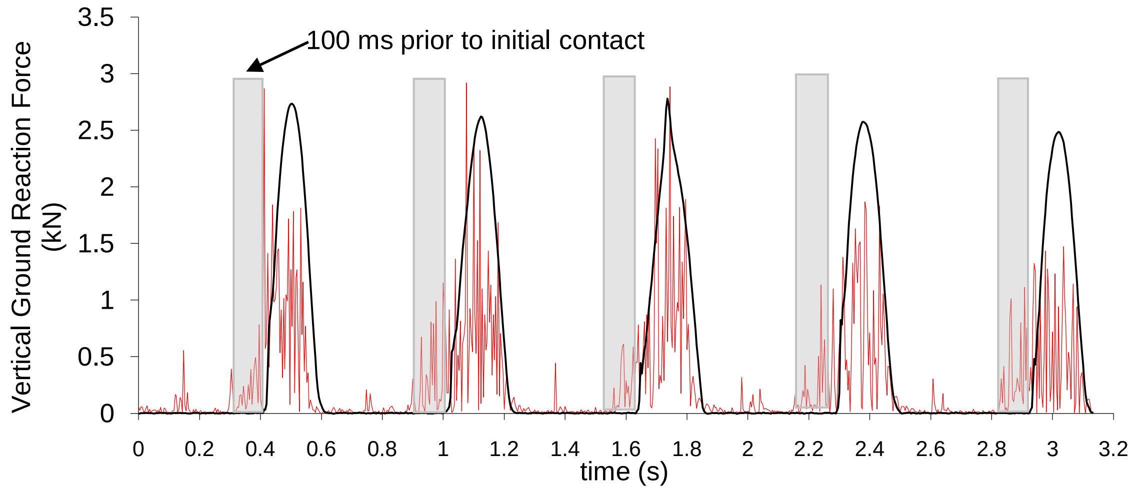


Figure 3.3.12 An example from the current study of left medial gastrocnemius muscle surface electromyography (full-wave rectified and band-pass filtered 20-500Hz) (red) and vertical ground reaction force recording (black) of five consecutive hops during double-leg hopping on a flat surface at a self-selected pace. Pre-loading activation amplitude defined as the sum of values over a 100ms window preceding initial contact (grey shaded area).

3.3.6.v - Normalised Peak Activation Amplitude

The peak EMG signal amplitude for DL hopping on a flat surface was used as a within participant normalisation procedure. Thus, the peak activation amplitude for each muscle in each trial was divided by the peak activation amplitude value for the same muscle during DL hopping on a flat surface at a self-selected pace and multiplied by 100 to yield the percentage normalised peak amplitude. Normalised peak amplitude was calculated for the purpose of graphing results across hopping conditions for each muscle.

3.3.6.v - Normalised Pre-Loading Activation Amplitude of Medial Gastrocnemius

The peak EMG signal amplitude for DL hopping on a flat surface at a self-selected pace was used as a within participant normalisation procedure. Thus, the pre-loading activation amplitude value of MG in each trial was divided by the peak value for the same muscle during DL hopping on a flat surface at a self-selected pace and multiplied by 100 to yield percentage normalised pre-loading activation amplitude. Statistical analysis was performed on the normalised pre-loading activation amplitude values for each trial across hopping conditions for MG on the left and right sides. This procedure of normalisation is consistent with studies that have examined EMG normalisation techniques using a dynamic task (Allison, et al., 1993; Bolgla & Uhl, 2007; Yang & Winter, 1984).

3.3.6.v - Co-activation Ratio

The agonist-antagonist co-activation ratio (CR) was calculated at the ankle and the knee as in previous studies (Billot, Simoneau, Van Hoecke, & Martin, 2010; Hobara, et al., 2010a; Kellis & Kouvelioti, 2009). At the ankle the CR was calculated as the quotient of the normalised peak activation amplitude between soleus and TA (CR_a) and between the MG and TA (CR_b). At the knee the CR was calculated as the quotient of the normalised peak activation amplitude between the VL and BF (CR_c). The CR between the prime mover at the knee and ankle was also calculated as the quotient between the VL and MG (CR_d). The MG is a two joint muscle and acts as a knee flexor and ankle extensor (Signorile, Applegate, Duque, Cole, & Zink, 2002).

The method used is consistent with a number studies investigating co-activation ratios at the knee and ankle (Billot, et al., 2010; Hagood, Solomonow, Baratta, Zhou, & D'Ambrosia, 1990; Hortobagyi & De Vita, 2000; Kellis, et al., 2009; Padua, et al., 2005; Palmieri-Smith, G., Ashton-Miller, & Wojtys, 2009). In the current study each muscle was treated independently and not grouped as a part of a muscle group. This specifically refers to the role of MG that acts in synergy with the soleus to extend the ankle (Signorile, et al., 2002). The evidence suggests that the role of each muscle varies independently to each other (Signorile, et al., 2002; Tamaki, Kitada, Akamine, Sakou, & Kurata, 1997) and specifically relates to the MG that changes activity based at either joint (knee or ankle) or either antagonist at each joint (TA or VL). This change in function may be different in nature as agonist activity of the soleus at the ankle joint or BF at the knee joint. Hence, the CR was treated as a function of a change in individual muscle patterning rather than one that reflected the net change at a joint.

Moritani et al. (1991) demonstrated a difference in muscle activation levels between MG and soleus based on the phase of contraction (pre-landing activation and eccentric phases) and illustrates the different behaviour within ankle extensors during preferred frequency and maximal height hopping with the MG having greater levels of activation in both phases than soleus (Moritani, et al., 1991). This was particularly important in evaluating hopping that is performed at relatively high joint angular velocities and requires high loading capacities of the MTU. Thus, the level of activation of each MTU may change differently to a synergist as the muscles vary significantly in muscle CSA and SEC. This may reflect significant differences in the ability to store and utilise elastic energy. In this case, assessment of the summed contribution of the contractile components that is largely reflected by muscle activation levels may not be sensitive to the changes of individual muscle and their effect on levels of co-contraction.

The process of calculating the CR used is described:

1. Normalisation of peak activation amplitude - Divided the peak activation amplitude values of each muscle during a hopping condition by the peak activation amplitude during DL hopping on the flat surface at a self-selected pace. This normalised the peak activation amplitude for each muscle to a reference task.

2. Divide the normalised peak activation amplitude for one muscle by the second muscle for each ratio (CR_a, CR_b, CR_c, CR_d). Thus, during DL hopping on a flat surface at a self-selected pace, the numerator and denominator for normalised peak activation were always one and therefore the CR was always one. As CR are calculated with reference to DL hopping on a flat surface at a self-selected pace that was seen to represent the reference CR of 1, any significant change in CR must be evaluated with respect to this reference task rather than seen as an increase in or decrease in peak activation amplitude. For example a change in CR may be due to a change in either the numerator or denominator or both values, however this does not mean that there was a concomitant and parallel change in peak activation amplitude of the muscle(s).

3.3.7 Data Analyses

3.3.7.i Inter-Side Comparisons

3.3.7.i - Kinetic and Joint Derived Variables

Inter-side comparisons were performed for kinetic and kinematic derived variables including: hopping frequency, vertical displacement of COM, vGRF, joint excursion, joint moment and joint stiffness. This was performed by using a paired t-test with statistical significance accepted at $p < 0.01$ (PASW Statistics, version 18). A more conservative alpha level was used as there were a number of comparisons made. If there was no statistically significant inter-side difference in the derived variable then the mean between left and right sides was calculated and used for analysis across hopping conditions. The observed power for each statistical test was calculated (G*Power 3.1.2, available as freeware on http://www.psycho.uni-dusseldorf.de).

3.3.7.i - Electromyography Derived Variables

Inter-side comparisons were performed for EMG data on the following two derived variables.
1.Change score for normalised peak activation amplitude – This was calculated as the difference between normalised peak activation amplitude for a condition of hopping minus the normalised peak activation amplitude during DL hopping on a flat surface at a self-selected pace and expressed as a percentage score.

2. Normalised pre-loading activation amplitude for MG. Multiple paired t-tests were performed on the change score for peak activation amplitude for each muscle (TA, soleus, MG, BF, VL and GM) and the pre-loading activation amplitude for MG to detect any inter-side differences with significance accepted at $p < 0.01$ (PASW Statistics, version 18). A more conservative alpha level was used as there were a number of comparisons made. If there was no statistically significant inter-side difference in the derived variable then the mean between left and right sides was calculated and used for analysis across hopping conditions. The observed power for each statistical test was calculated using G*Power 3.1.2 (available as freeware on http://www.psycho.uni-dusseldorf.de).

3.3.7.ii Between Hopping Conditions

3.3.7.ii - Kinetic, Joint and Electromyography Derived Variables

A repeated 3-way ANOVA with factors for hopping task (DL and AL), hopping surface (flat and decline) and hopping pace (self-selected and forced pace) was performed (PASW Statistics, version 18). An alpha level of 0.05 was set as a threshold for statistical significance and in the case of multiple analyses Bonferroni correction was performed. This statistical test was used to compare across hopping conditions for derived variables including: hopping frequency, vertical height displacement of COM during braking phase, vGRF, vertical leg stiffness, joint excursion, joint moment, joint stiffness, joint work, left peak activation amplitudes (TA, soleus, MG, BF, VL and GM), right peak activation amplitudes (TA, soleus, MG, BF, VL and GM), left normalised MG pre-loading activation amplitudes and right MG normalised pre-loading activation amplitudes.

If there was a significant main effect, multiple paired t-tests were performed between each pair of derived variables for the single factor with Bonferroni correction, calculated as the alpha level divided by the product of the number of levels for each factor with no main effect (PASW Statistics, version 18). The calculated alpha levels were set *a priori* for post hoc analysis of significant main effect(s) (Appendix K). The observed power for each post hoc statistical test was calculated using G*Power 3.1.2 (available as freeware on http://www.psycho.uni-dusseldorf.de)).

Figure 3.3.13 illustrates the process of statistical analysis based on whether there was significant 3-way or 2-way interaction(s).

One-way ANOVA was performed for comparison of left and right co-activation ratios (CR_a, CR_b, CR_c and CR_d) across hopping conditions with Bonferroni correction made for multiple analyses. Pairwise comparisons are reported when there was a significant difference in the derived variables ($p < 0.05$) (PASW Statistics, version 18).

3.3.7.iii Comparison of Reference and Field Test Methods - Vertical Leg Stiffness

Vertical leg stiffness was calculated using the two methods described and a linear regression analysis performed to compare the two sets of values for vertical stiffness calculated for each of the eight hopping conditions. This study is detailed in chapter 4.

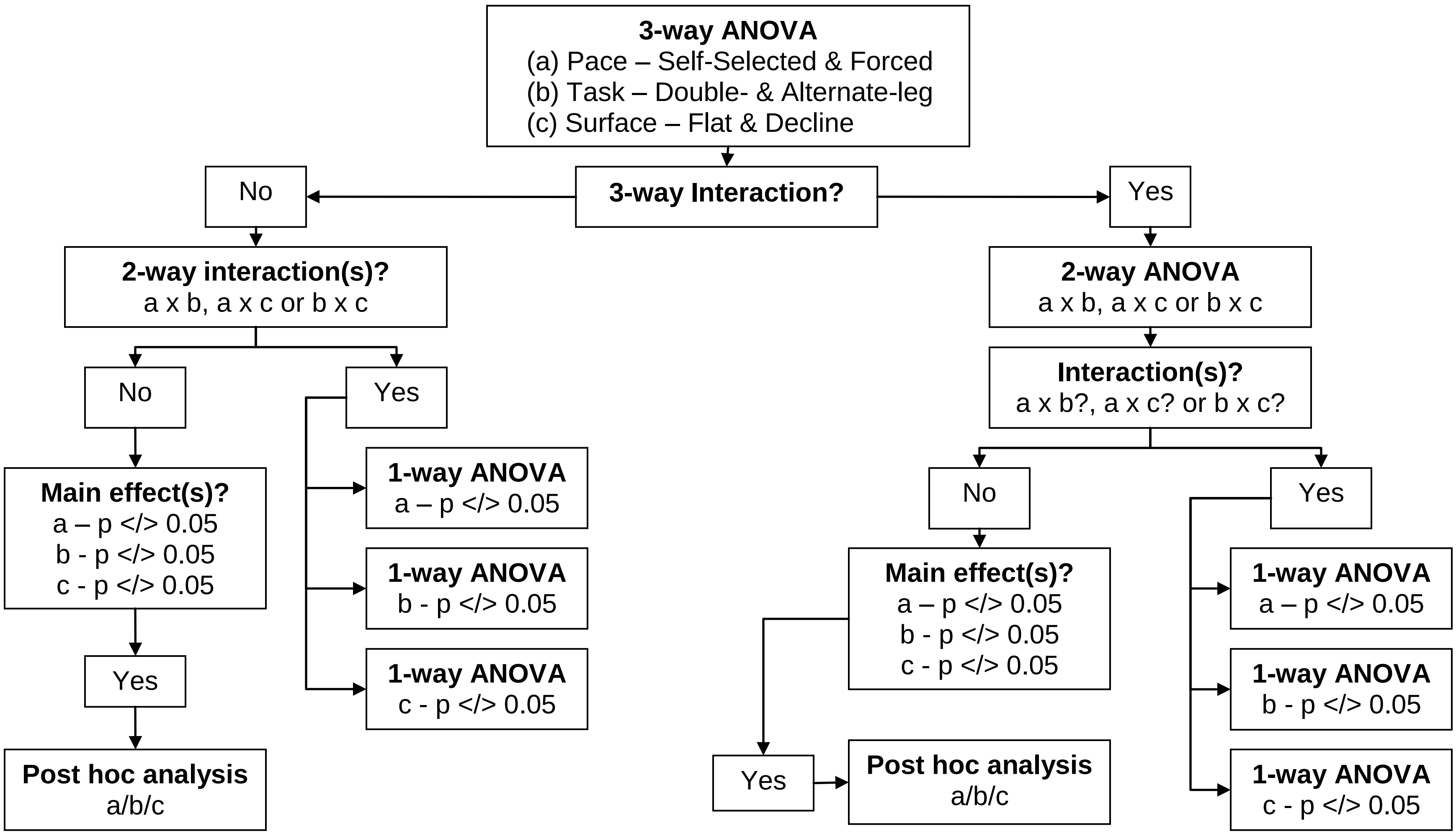


Figure 3.3.13 Flow chart demonstrating the process of statistical analysis for a 3-way ANOVA with three factors (a, b and c) with two levels for each factor.

3.4 Results

3.4.1 Comparisons between Left and Right Sides

This investigation did not detect any systematic inter-side differences ($p > 0.01$) in the derived variables for hopping frequency, peak vGRF, vertical height displacement, joint excursion, joint moment, joint stiffness, change scores for normalised peak activation amplitude and normalised pre-loading activation amplitude. A total of 152 inter-side comparisons were made for 19 derived variables. The results of these statistical comparisons are presented in Appendix H.

3.4.2 Comparisons between Hopping Conditions

The mean of left and right sides was calculated for hopping frequency, temporal, kinetic and joint variables including joint mechanical work done.

The results for comparisons of derived variables for hopping frequency, kinetics, kinematics and electromyography across hopping conditions, is divided into three main sections defined by the derived variables. Descriptive data for the derived variables in each section are presented at the start of each of these sections with statistical results detailed in Appendix I for hopping frequency and kinetic variables, Appendix J for joint variables and Appendix K for EMG variables.

3.4.2.i - Hopping Frequency and Kinetics – Between Hopping Condition Comparisons

Descriptive results of hopping frequency, temporal and kinetic data during hopping across conditions are presented in Table 3.4.1.

Table 3.4.1 Descriptive data (mean (SD)) for hopping frequency, temporal and kinetic variables.

	Self-Selected Pace				Forced Pace (1.8 Hz)			
	DL Flat	DL Dec	AL Flat	AL Dec	DL Flat	DL Dec	AL Flat	AL Dec
Hopping Frequency for the COM (Hz)	1.97 (0.23)	2.04 (0.18)	2.40 (0.19)	2.44 (0.27)	1.84 (0.07)	1.83 (0.04)	1.86 (0.06)	1.84 (0.07)
Flight time of the COM (s)	0.25 (0.06)	0.18 (0.05)	0.06 (0.03)	0.06 (0.04)	0.29 (0.04)	0.25 (0.05)	0.15 (0.06)	0.13 (0.06)
Contact time of the COM (s)	0.27 (0.05)	0.28 (0.05)	0.37 (0.05)	0.39 (0.09)	0.26 (0.04)	0.29 (0.04)	0.39 (0.056)	0.41 (0.06)
Braking phase (s)	0.13 (0.03)	0.16 (0.03)	0.18 (0.04)	0.19 (0.06)	0.12 (0.03)	0.15 (0.03)	0.2 (0.04)	0.21 (0.04)
z_{flight} of COM (cm)	7.90 (4.65)	4.46 (2.40)	0.55 (0.47)	0.46 (0.59)	10.48 (3.09)	8.22 (3.09)	3.08 (2.10)	2.41 (1.78)
$z_{braking}$ COM (cm)	8.02 (1.75)	6.58 (1.40)	2.49 (1.40)	2.64 (1.29)	8.98 (0.81)	8.95 (0.86)	6.79 (1.90)	6.17 (2.21)
Peak $vGRF_{BM}$	3.87 (0.73)	4.18 (0.65)	2.37 (0.32)	2.03 (0.35)	2.97 (0.45)	3.23 (0.49)	2.49 (0.375)	2.09 (0.32)
k_{LBM} ($kN.m^{-1}.kg^{-1}$)	0.49 (0.11)	0.45 (0.07)	1.28 (0.75)	0.96 (0.41)	0.46 (0.10)	0.35 (0.04)	0.39 (0.13)	0.40 (0.23)
Impulse ($\Delta F.t$)	910.59 (145.61)	743.03 (140.59)	760.83 (93.79)	681.20 (78.25)	936.57 (56.95)	829.94 (64.58)	848.58 (61.56)	755.14 (44.11)

DL – Double-Leg hopping; AL – Alternate-Leg hopping; Dec – Decline Surface; L – Left; R – Right; z – maximum vertical height displacement; COM – Centre of Mass; Peak $vGRF_{BM}$ – Peak vertical Ground Reaction Force normalised to body mass (BM) (kg); k_{LBM} – Vertical Leg Stiffness normalised to body mass (BM); ΔF – Change in force in braking phase (N); t – time (s)

3.4.2.i - Hopping Frequency

There was no difference in hopping frequency in hopping conditions at a forced pace. Hopping frequency was significantly greater during AL hopping on both surfaces at a self-selected pace than DL hopping on both surfaces at a self-selected pace, during DL hopping on a flat surface at a forced pace and AL hopping on both surfaces at a forced pace ($p < 0.01$) (Figure 3.4.1) (Appendix I: Figure I.1).

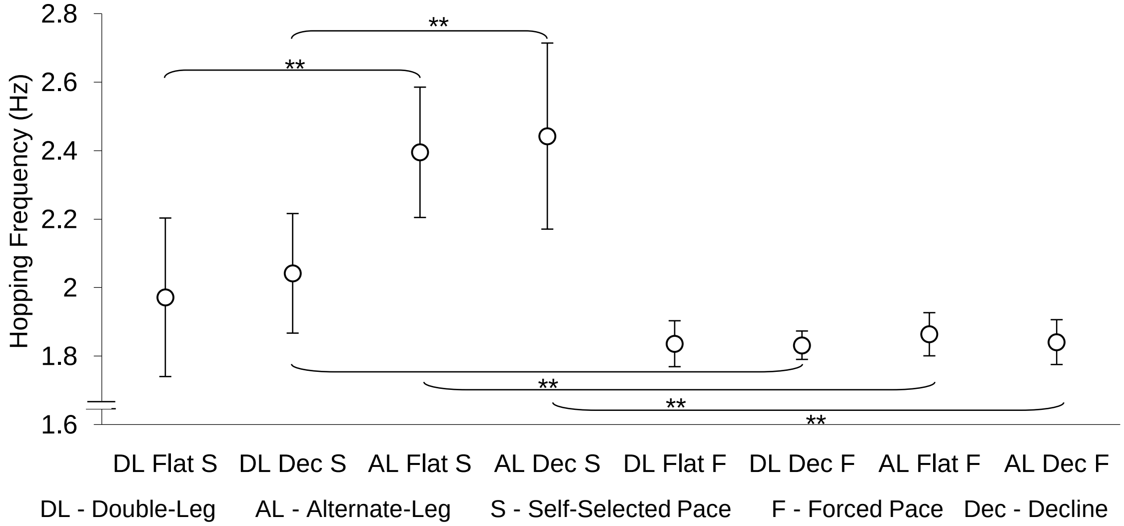


Figure 3.4.1 Hopping frequency (Hz) (mean SD) during alternate-leg (AL) hopping on both surfaces is greater than double-leg (DL) hopping on both surfaces at a self-selected pace, DL hopping on the flat at a forced pace and AL hopping on both surfaces at a forced pace (p < 0.01).**

3.4.2.i - Maximum Vertical Height Displacement in Flight Phase

The maximum vertical displacement in flight phase (z_f) was significantly greater during DL than AL hopping on both surfaces and at both paces ($p < 0.01$). The z_f during DL hopping at a self selected pace was significantly greater during hopping on a flat than decline surface ($p = 0.03$). The z_f was significantly greater during forced than self selected pace hopping during DL hopping on a decline surface ($p = 0.01$), AL hopping on a flat surface ($p = 0.002$) and AL hopping on a decline surface ($p = 0.003$) (Figure 3.4.2) (Appendix I: Figures I.2).

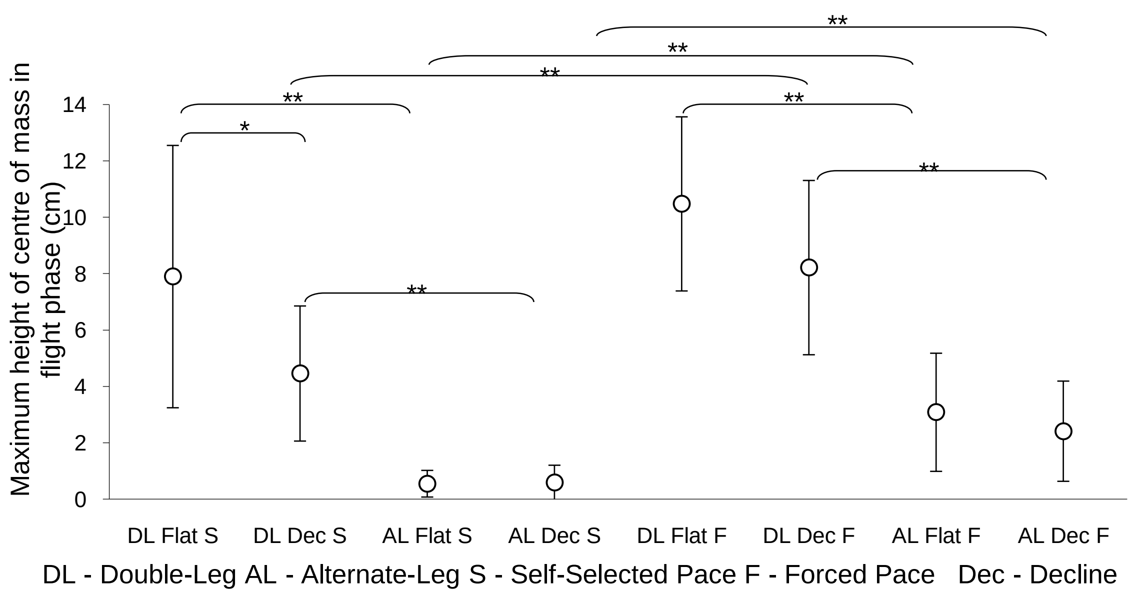


Figure 3.4.2 The vertical height displacement (cm) (z_f) (mean SD) in flight phase was greater during double-leg (DL) than alternate-leg (AL) hopping on both surfaces and at both paces ($p < 0.01$). The z_f during DL hopping at a self selected pace was greater during hopping on a flat than decline surface (*$p < 0.05$). The z_f was greater during forced than self selected pace hopping during DL hopping on a decline surface and AL hopping on both surfaces (**$p < 0.01$).**

3.4.2.i - Peak Vertical Ground Reaction Force

Normalised peak vGRF was significantly lower when hopping on decline compared to flat surface for DL and AL hopping at both paces ($p < 0.01$) (Figure 3.4.3) (Appendix I: Figure I.3). Peak vGRF was significantly lower for AL compared to DL hopping on both surfaces and at both paces (Figure 3.4.3) (Appendix I: Figure I.3).

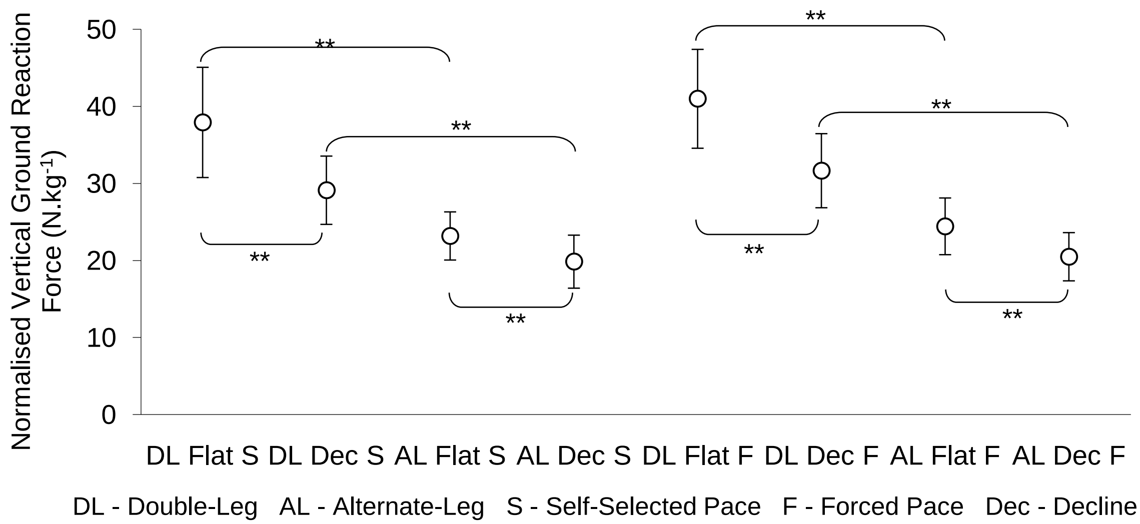


Figure 3.4.3 Normalised peak vertical ground reaction force ($N.kg^{-1}$) (mean SD) at both hopping paces was significantly lower during alternate-leg (AL) compared to double-leg (DL) hopping on both surfaces ($p < 0.01$) and lower during decline compared to flat surface hopping during both DL and AL hopping tasks (**$p < 0.01$).**

3.4.2.i - Impulse

Impulse was significantly greater when hopping on a flat compared to the decline surface for both tasks at both paces ($p < 0.01$) (Figure 3.4.4) (Appendix I: Figure I.4). Impulse was significantly greater during DL than AL hopping on a flat surface at both paces ($p < 0.01$) and on a decline surface at a forced pace ($p < 0.01$) (Figure 3.4.4) (Appendix I: Figure I.4). Impulse was significantly greater at a forced pace compared to a self-selected pace for AL hopping on both surfaces ($p < 0.01$) and for DL hopping on a decline surface ($p < 0.01$) (Figure 3.4.4) (Appendix I: Figure I.4).

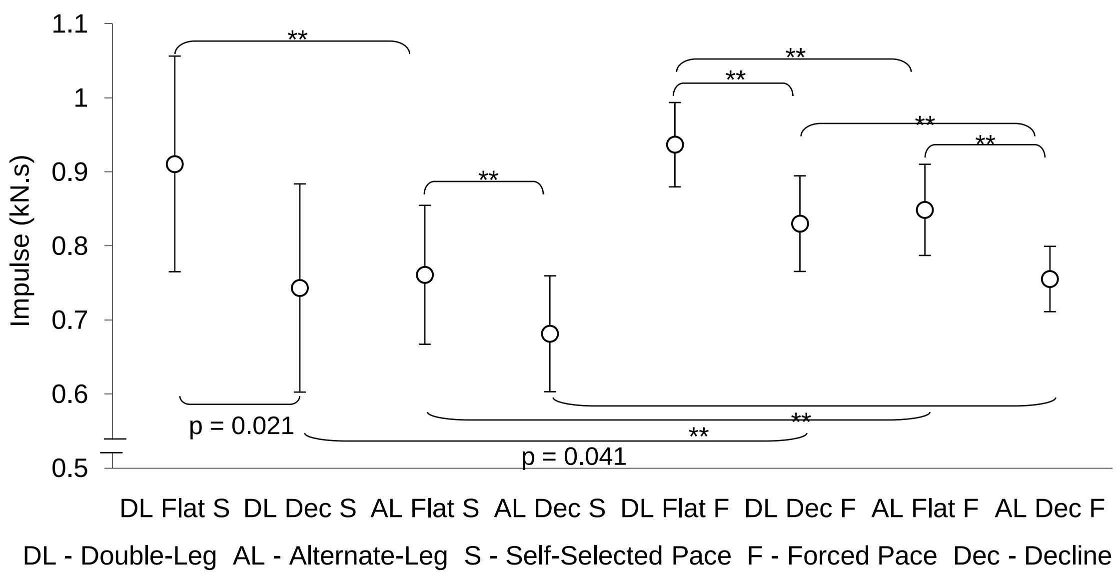


Figure 3.4.4 Impulse (kN.s) (mean SD) was significantly greater during flat compared to decline surface hopping for both double-leg (DL) and alternate-leg (AL) hopping at a self-selected (S) and forced (F) pace of hopping (p < 0.01). Impulse was significantly greater during DL than AL hopping on a flat surface at a S pace and on both surfaces at a F pace (**). Impulse was significantly greater during F than S pace hopping during DL hopping on decline surface and during AL hopping on both surfaces (**p < 0.01).**

3.4.2.i - Vertical Height Displacement in Braking Phase

Vertical height displacement was significantly greater during DL than AL hopping on both surfaces and at both paces ($p < 0.05$) (Figure 3.4.5) (Appendix I: Figure I.5). Vertical height displacement was significantly greater at a forced compared to self-selected pace during AL hopping on both surfaces ($p < 0.01$) and during DL hopping on a decline surface ($p < 0.01$) (Figure 3.4.5) (Appendix I: Figure I.5).

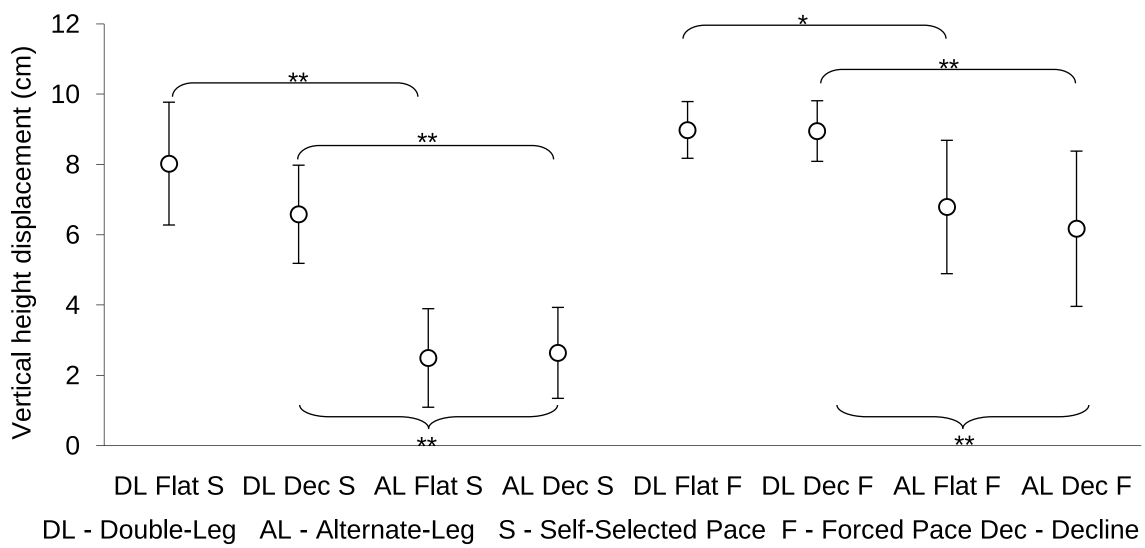


Figure 3.4.5 Vertical height displacement (cm) (mean SD) in braking phase was significantly greater during double-leg (DL) than alternate-leg (AL) hopping on both surfaces and at both paces (*p < 0.05 & **p < 0.01). Vertical height displacement was significantly greater during forced compared to self-selected pace hopping during AL hopping on both surfaces and during DL hopping on a decline surface (p < 0.01).**

3.4.2.i - Vertical Leg Stiffness

Vertical leg stiffness was significantly greater during AL than DL hopping at a self-selected pace on both surfaces ($p < 0.05$) (Figure 3.4.6) (Appendix I: Figure I.6). Vertical leg stiffness was significantly lower at a forced compared to self-selected pace during DL hopping on a decline surface and during AL hopping on both surfaces ($p < 0.01$) (Figure 3.4.6) (Appendix I: Figure I.6).

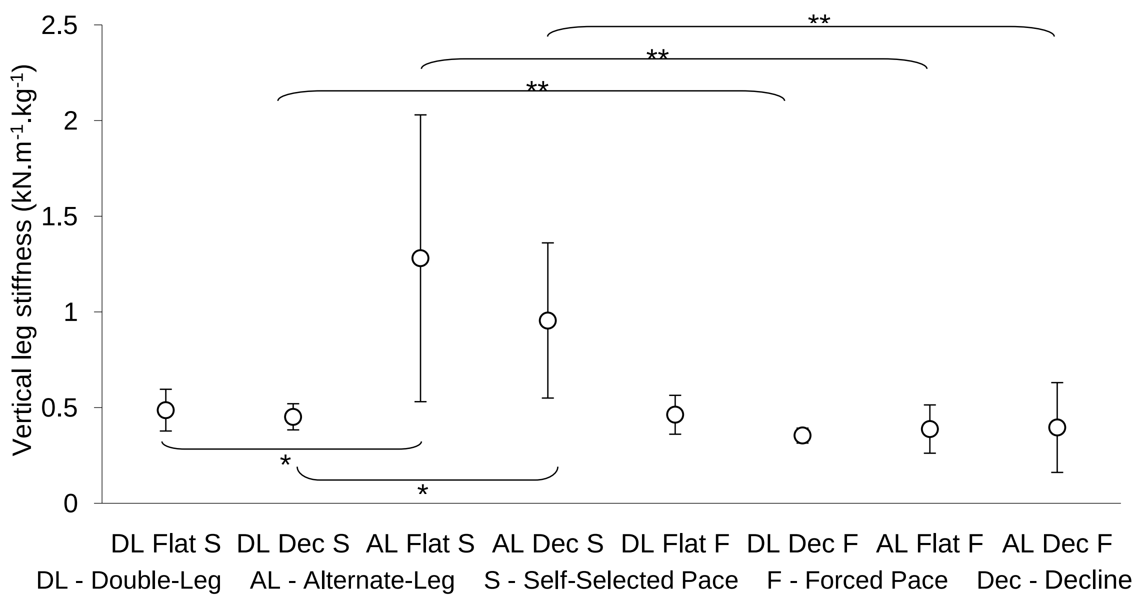


Figure 3.4.6 Vertical leg stiffness ($kN.m^{-1}.kg^{-1}$) (mean SD) was significantly greater during alternate-leg (AL) than double-leg (AL) hopping at a self-selected pace on both surfaces (*$p < 0.05$). Vertical leg stiffness was significantly lower during forced compared to self-selected pace hopping during DL hopping on a decline surface and AL hopping on both surfaces ($p < 0.01$).**

Summary of Findings – Hopping Frequency and Kinetics

- No systematic differences between left and right sides were detected.
- At a self-selected pace participants reduced their hopping frequency during AL compared to DL hopping on both surfaces (flat and decline). Hopping surface did not affect hopping frequency.
- There was greater impulse on the flat than decline surfaces and during DL than AL hopping.
- Peak vGRF was greater for DL than AL. Peak vGRF was greater on the flat than decline surface for hopping.

- Vertical height displacement was greater during DL hopping compared to AL hopping at both pace and on both surfaces.
- Vertical leg stiffness was similar in hopping conditions when hopping frequency was forced at 1.8Hz. Vertical leg stiffness was greater during AL than DL hopping on both surfaces at a self-selected pace. These results indicate that vertical leg stiffness is primarily governed by the hopping frequency rather than the surface or task.

3.4.2.ii Joint Variables – Between Hopping Condition Comparisons

3.4.2.ii - Joint Excursion

Descriptive results are shown in Table 3.4.2 for joint excursions at the ankle, knee and hip joints during braking phase. Statistical differences are shown in subsequent figures.

Table 3.4.2 Joint excursion (degrees) (mean (SD)) for ankle, knee and hip joints during braking phase on the left and right sides under each hopping condition.

Hopping Condition	Ankle	Knee	Hip
Double-Leg Flat Self-Selected Pace	39.7 (6. 7)	20.6 (6.5)	7.3 (5.1)
Double-Leg Decline Self-Selected Pace	33.9 (6.9)	28.6 (5.9)	13.0 (4.7)
Alternate-Leg Flat Self-Selected Pace	31.3 (4. 8)	8.5 (4.0)	-0.1 (3.9)
Alternate-Leg Decline Self-Selected Pace	25.6 (5.2)	13.5 (5.2)	3.0 (3.4)
Double-Leg Flat Forced Pace	40.0 (5.8)	21.8 (6.0)	7.6 (3.5)
Double-Leg Decline Forced Pace	38.8 (3.9)	32.8 (4.2)	13.5 (4.3)
Alternate-Leg Flat Forced Pace	43.3 (4.8)	20.5 (3.8)	7.1 (3.5)
Alternate-Leg Decline Forced Pace	36.0 (3.2)	26.2 (4.3)	11.2 (3.8)

Ankle excursion was significantly lower when hopping on a decline compared to flat surface during DL at a self-selected pace ($p = 0.01$) and during AL hopping at both paces ($p < 0.01$) (Figure 3.4.7) (Appendix J: Figure J.1). Ankle excursion was significantly lower during AL compared to DL hopping on the flat ($p = 0.008$) and decline ($p = 0.03$) surface at a self-selected pace (Figure 3.4.7) (Appendix J: Figure J.1). Ankle excursion was significantly greater during forced compared to the self-selected pace of hopping during DL hopping on a decline surface ($p = 0.02$) and during AL hopping on both surfaces ($p < 0.01$) (Figure 3.4.7) (Appendix J: Figure J.1).

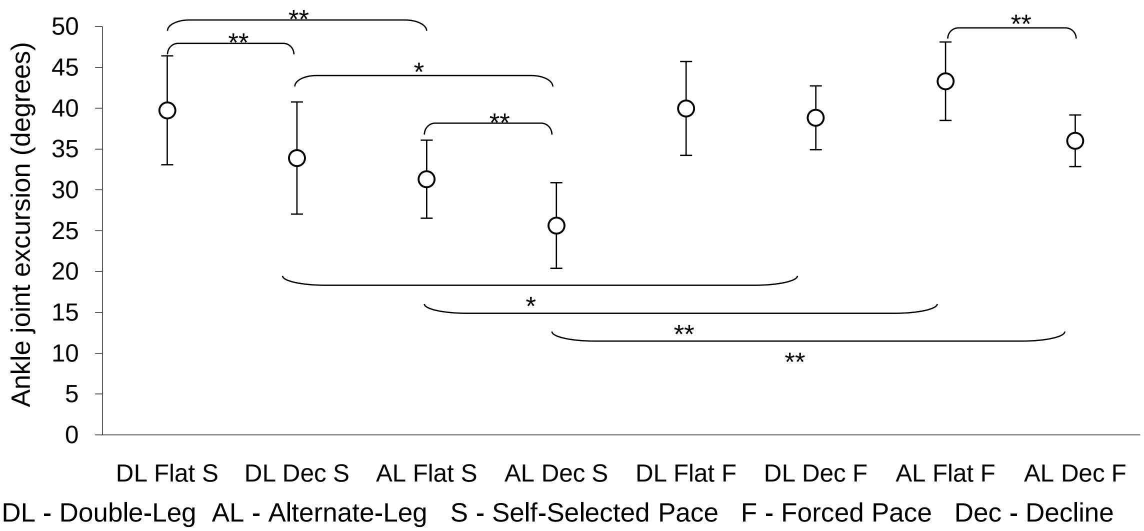


Figure 3.4.7 Ankle joint excursion (degrees) (mean SD) was significantly lower during hopping on a decline compared to flat surface during double-leg (DL) and alternate-leg (AL) hopping at a self-selected (S) pace and during AL hopping at a forced pace (F) (p < 0.01). Ankle excursion was lower during AL than DL hopping at a self-selected (S) pace on the flat (**p < 0.01) and decline surface (**p < 0.01). Ankle joint excursion was greater during forced than S pace hopping during DL hopping on a decline surface (*) and during AL hopping on both surfaces (**p < 0.01).**

The findings for knee excursion were in contrast to ankle excursion. Knee excursion was significantly greater when hopping on a decline compared to flat surface during DL hopping at both paces ($p < 0.01$) and during AL hopping at a forced pace ($p = 0.003$) (Figure 3.4.8) (Appendix J: Figure J.2). Knee joint excursion was significantly lower during AL compared to DL hopping on both surfaces at a self-selected (S) pace ($p < 0.01$) and on both surfaces at a forced (F) pace ($p < 0.01$) (Figure 3.4.8) (Appendix J: Figure J.2). Knee excursion was significantly greater during forced compared to the self-selected pace of hopping during AL hopping on both surfaces ($p < 0.01$) (Figure 3.4.8) (Appendix J: Figure J.2).

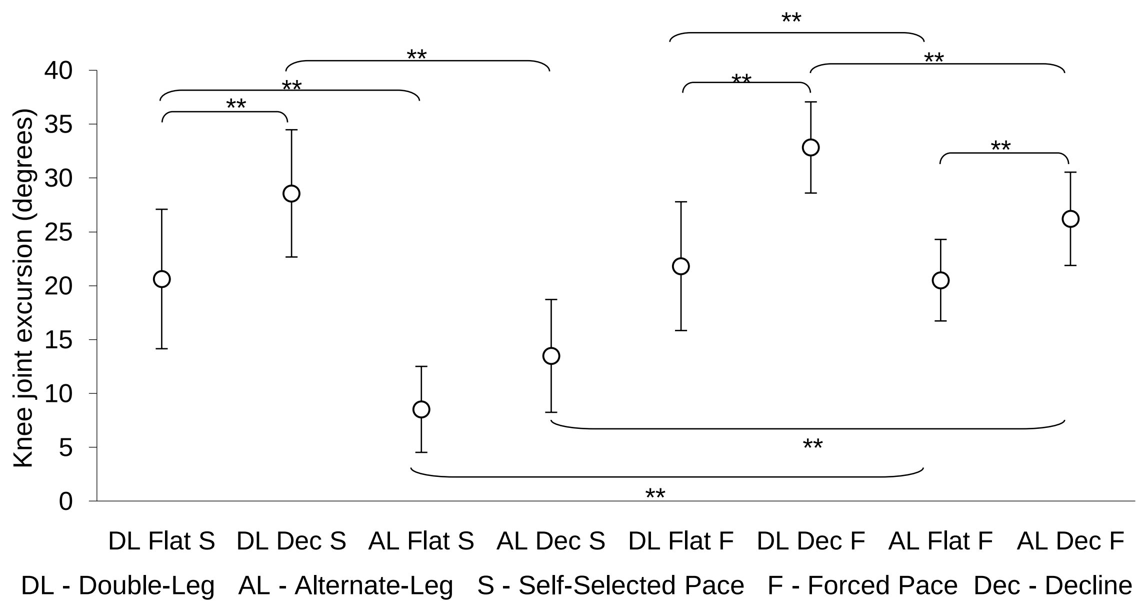


Figure 3.4.8 Knee joint excursion (degrees) (mean SD) was significantly greater when hopping on a decline compared to flat surface during double-leg (DL) hopping at both paces (p < 0.01) and during alternate-leg (AL) hopping at a forced pace (**p < 0.01). Knee joint excursion was significantly lower during AL compared to DL hopping on both surfaces at a self-selected (S) pace (**p < 0.01) and on both surfaces at a forced (F) pace (**p < 0.01). Knee excursion was significantly greater during F compared to S pace hopping during AL hopping on both surfaces (**p < 0.01).**

Similar to the knee, hip joint excursion was greater during decline surface than flat surface hopping detected on both surfaces at a self-selected pace and on the decline surface at a forced pace of hopping. Hip joint excursion was significantly greater when hopping on a decline compared to flat surface during DL hopping at a both paces ($p < 0.01$) and during AL hopping at a forced pace ($p = 0.002$) (Figure 3.4.9) (Appendix J: Figure J.3). Hip joint excursion was significantly greater during DL than AL hopping on the flat surface at a self-selected pace ($p < 0.01$) on the decline surface at a self-selected pace ($p < 0.01$) and on the decline surface at a forced pace ($p = 0.03$) (Figure 3.4.9) (Appendix J, Figure J.3). Hip joint excursion was significantly greater during forced than self-selected pace hopping during AL hopping on both surfaces ($p < 0.01$) (Figure 3.4.9) (Appendix J: Figure J.3).

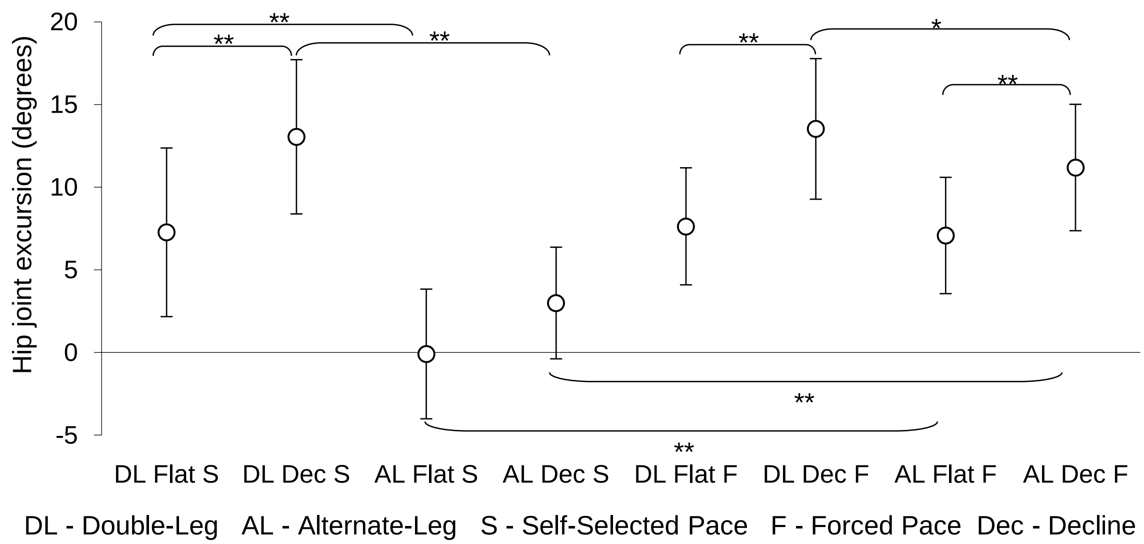


Figure 3.4.9 Hip joint excursion (degrees) (mean SD) was significantly greater during hopping on a decline surface compared to flat surface during double-leg (DL) hopping at both paces (p < 0.01) and during alternate-leg (AL) hopping at a forced pace (**p < 0.01). Hip joint excursion was significantly greater during DL than AL hopping on both surfaces a self-selected pace (**p < 0.01) and on the decline surface at a forced pace (*p < 0.01). Hip joint excursion was significantly greater during hopping at a force compared to a self-selected pace during AL hopping on both surfaces (**p < 0.01).**

The high CV for hip excursion ranging from -4073.0 to 113.1% was due to the small mean values that ranged from mean of -0.1º (SD) (3.9) during AL hopping on a flat surface at a self-selected pace to 13.51º (4.3) during DL hopping on a decline surface at a forced pace. The minimal hip joint excursions were due to the nature of the task in that hopping was performed avoiding motion in the saggital and coronal planes. Thus, hip joint excursions were minimised by the participants to avoid displacing the COM that lies approximately at the level of the hips. Negative values represent hip extension and positive values represent flexion.

3.4.2.ii - Joint Angular Velocity

Descriptive results are shown in Table 3.4.3 for joint angular velocity at the ankle, knee and hip joints during braking phase.

Table 3.4.3 Joint angular velocity (degrees/second) (mean (SD)) for ankle, knee and hip joints during braking phase on the left and right sides during each hopping condition.

Hopping Condition	Ankle	Knee	Hip
DL Flat Self-Selected Pace	296.6 (47.9)	149.7 (36.3)	50.43 (34.1)
DL Decline Self-Selected Pace	210.4 (50. 9)	175.4 (25.1)	81.32 (29.2)
AL Flat Self-Selected Pace	172.4 (44.2)	46.3 (24.9)	-1.10 (21.8)
AL Decline Self-Selected Pace	137.7 (47.4)	72.5 (29.4)	15.86 (17.7)
DL Flat Forced Pace	324.4 (42.6)	172.5 (25.1)	59.19 (24.0)
DL Decline Forced Pace	247.0 (31.2)	205.6 (26.7)	82.84 (17.3)
AL Flat Forced Pace	222.1 (47.5)	105.9 (31.4)	36.35 (19.6)
AL Decline Forced Pace	179.4 (45.2)	126. 5 (34.4)	53.13 (17.0)

DL – Double-Leg; AL – Alternate-Leg

3.4.2.ii – Normalised Joint Moment

Descriptive results are shown in Table 3.4.4 for change in normalised joint moments (from IC to peak vGRF) at the ankle, knee and hip joints during braking phase.

Table 3.4.4 Change in normalised joint moment from initial contact to peak vertical ground reaction force ($Nm.kg^{-1}$) (mean (SD)) for ankle, knee and hip joints during braking phase across hopping conditions.

Hopping Condition	Ankle	Knee	Hip
DL Flat Self-Selected Pace	3.02 (0.63)	1.34 (0.58)	0.33 (0.71)
DL Decline Self-Selected Pace	2.14 (0.35)	1.58 (0.49)	0.82 (0.59)
AL Flat Self-Selected Pace	3.76 (0.54)	6.57 (0.70)	1.08 (0.42)
AL Decline Self-Selected Pace	2.58 (0.40)	1.16 (0.54)	0.97 (0.70)
DL Flat Forced Pace	3.38 (0.74)	1.66 (0.34)	0.70 (0.71)
DL Decline Forced Pace	2.20 (0.40)	1.86 (0.46)	0.83 (0.67)
AL Flat Forced Pace	3.85 (0.70)	1.47 (0.70)	1.60 (0.31)
AL Decline Forced Pace	2.69 (0.49)	1.63 (0.56)	1.61 (0.72)

DL – Double-Leg; AL – Alternate-Leg

The normalised ankle joint moment was significantly lower when hopping on a decline compared to a flat surface during both tasks at both paces ($p < 0.01$) (Figure 3.4.10) (Appendix J: Figure J.4). Normalised ankle joint moment was significantly lower during DL than AL hopping on a decline surface at a self-selected pace ($p = 0.004$) (Figure 3.4.10) (Appendix J: Figure J.4).

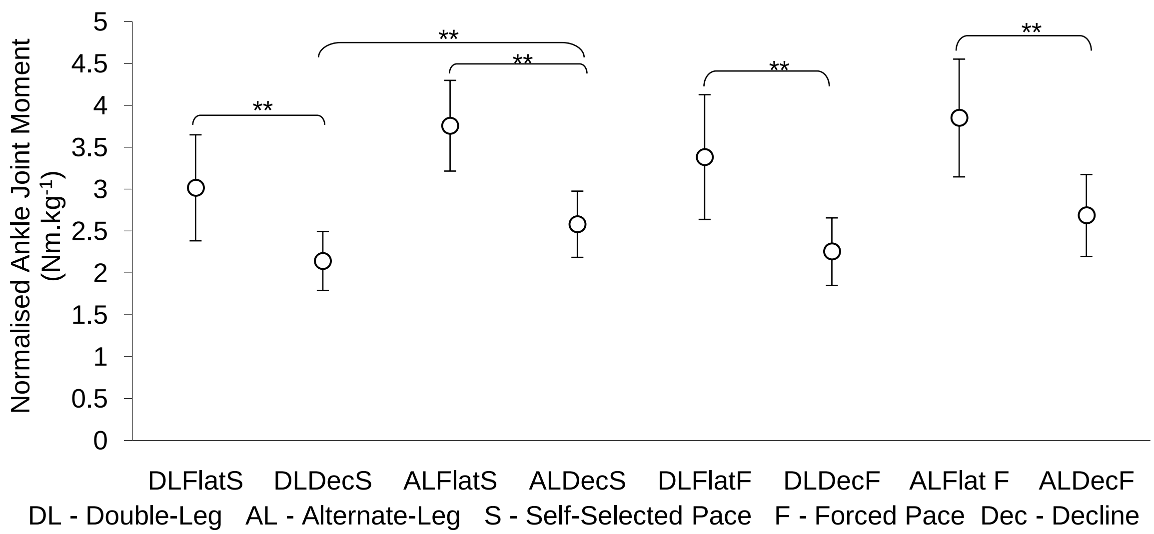


Figure 3.4.10 Normalised ankle joint moment ($Nm.kg^{-1}$) (mean SD) was significantly lower when hopping on a decline compared to flat surface during both hopping tasks and at both hopping paces ($p < 0.01$). Normalised ankle joint moment was lower during double-leg than alternate-leg hopping on a decline surface at a self-selected pace (**$p < 0.01$).**

Normalised knee joint moment was significantly lower during AL than DL hopping on the flat ($p = 0.005$) and decline ($p = 0.011$) surfaces (Figure 3.4.11) (Appendix J: Figure J.5). Normalised knee joint moment was significantly greater at a forced compared to a self-selected pace during AL hopping on both surfaces ($p < 0.01$) (Figure 3.4.11) (Appendix J: Figure J.5). Normalised knee joint moment was similar across hopping conditions at a forced pace ($p > 0.05$) (Figure 3.4.11) (Appendix J: Figure J.5).

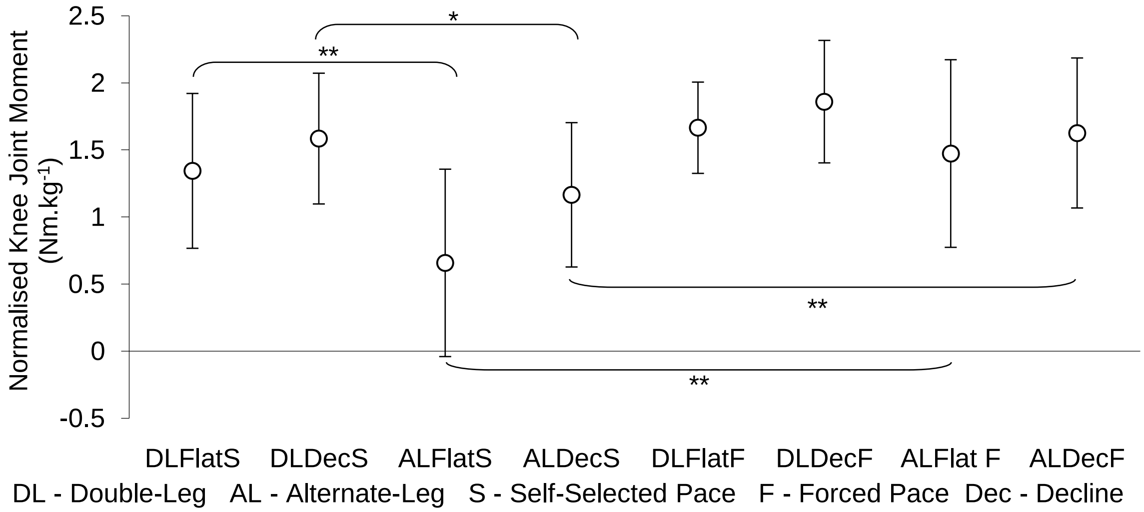


Figure 3.4.11 Normalised knee joint moment (Nm.kg^{-1}) (mean SD) was significantly lower during alternate-leg (AL) than double-leg hopping tasks on the flat (p < 0.01) and decline (*p < 0.05) surfaces at a self-selected pace. Normalised knee joint moment was significantly greater during forced compared to the self-selected pace hopping during AL hopping on both surfaces (**p < 0.01).**

Normalised hip joint moment was significantly greater when during AL than DL hopping on a flat surface at a self-selected (p = 0.007) and forced (p = 0.002) pace of hopping (Figure 3.4.12) (Appendix J: Figure J.6). Normalised hip joint moment was significantly greater during forced compared to self-selected pace of hopping during AL hopping on both surfaces (p < 0.01) (Figure 3.4.12) (Appendix J: Figure J.6).

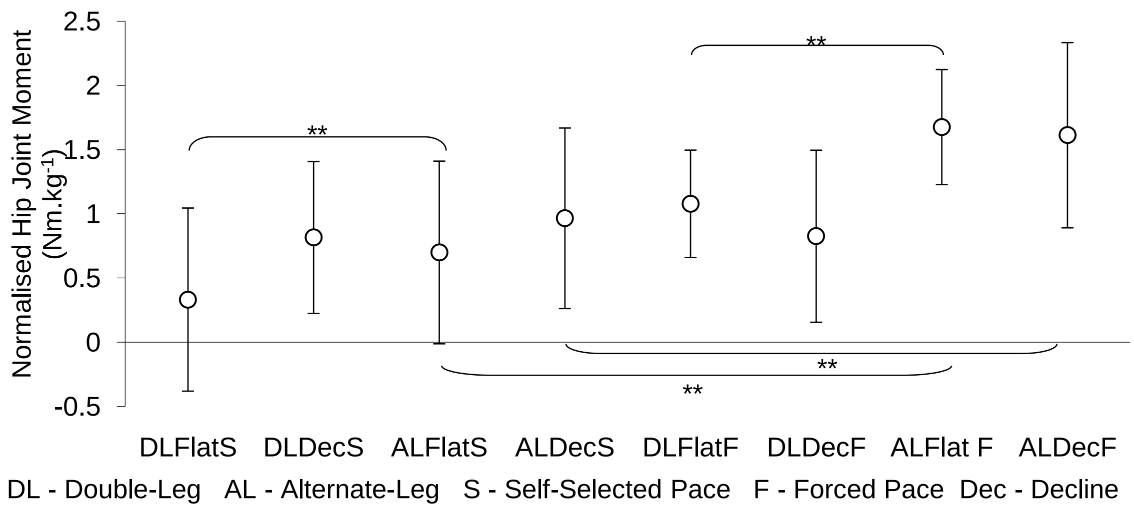


Figure 3.4.12 Normalised hip joint moment (Nm.kg^{-1}) (mean SD) was significantly greater during alternate-leg (AL) than double-leg hopping on a flat surface at a both paces (p < 0.01). Normalised hip joint moment was significantly greater during forced compared to the self-selected pace of hopping during AL hopping on both surfaces (**p < 0.01).**

3.4.2.ii – Normalised Joint Stiffness

Descriptive results are shown in Table 3.4.5 for normalised joint stiffness at the ankle, knee and hip joints during braking phase. A description of the analysis of the normalised ankle joint stiffness is included in Appendix J.7.Supplement that describes the data analysis for removal for the score of a single outlier during DL hopping on a decline surface at a self-selected pace.

Table 3.4.5 Normalised joint stiffness ($kNm.kg^{-1}.rad^{-1}$) (mean (SD)) for ankle, knee and hip joints during braking phase across hopping conditions.

Hopping Condition	Ankle	Knee	Hip
DL Flat Self-Selected Pace	4.50 (1.40)	3.94 (1.81)	6.47 (19.58)
DL Decline Self-Selected Pace	3.37 (0.06)†	3.29 (1.11)	3.81 (29.10)
AL Flat Self-Selected Pace	7.11 (0.91)	8.81 (11.23)	-0.90 (44.60)
AL Decline Self-Selected Pace	5.93 (0.88)	5.38 (2.24)	23.32 (90.20)
DL Flat Forced Pace	5.09 (1. 90)	4.64 (1.33)	8.57 (13.85)
DL Decline Forced Pace	3.35 (0.58)	3.34 (1.05)	3.41 (2.73)
AL Flat Forced Pace	5.24 (1.00)	4.19 (1.79)	12.16 (12.00)
AL Decline Forced Pace	4.29 (0.66)	3.54 (1.11)	8.65 (3.44)

DL – Double-Leg; AL – Alternate-Leg; S – Self-Selected Pace; F – Forced Pace; Dec – decline; † - n = 8

Normalised ankle joint stiffness was significantly lower during decline compared to flat surface hopping during both tasks at a self-selected pace ($p = 0.004$) (Figure 3.4.13) (Appendix J: Figure J.7). Normalised ankle joint stiffness was significantly lower during decline compared to flat surface hopping during both tasks at a forced pace ($p = 0.019$) (Figure 3.4.13) (Appendix J: Figure J.7). Normalised ankle joint stiffness was significantly greater during AL than DL hopping on both surfaces at a self-selected pace ($p \leq 0.001$) and on a decline surface at a forced pace ($p = 0.001$) (Figure 3.4.13). Normalised ankle joint stiffness was significantly lower at a forced compared to a self-selected pace during AL hopping on the flat ($p = 0.004$) and decline ($p = 0.003$) surfaces (Figure 3.4.13) (Appendix J: Figure J.7).

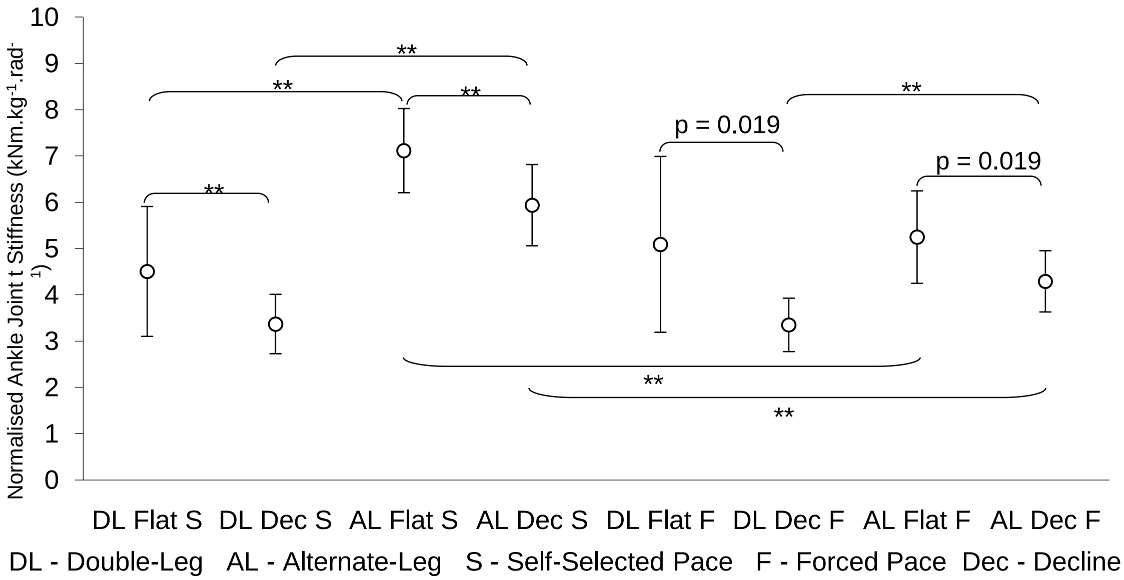


Figure 3.4.13 Normalised ankle joint stiffness (kNm.kg^{-1}.rad^{-1}) (mean SD) during braking phase was significantly lower hopping on a decline than flat surface during alternate-leg (AL) hopping and double-leg (DL)hopping at a self-selected pace (p < 0.01). Normalised ankle joint stiffness was close to statistical significance (p = 0.019) between surfaces for both tasks at a forced pace. Normalised ankle joint stiffness was greater during AL than DL hopping when hopping on a decline surface at both paces (**p < 0.01) and when hopping on a flat surface at a self-selected pace (**p < 0.01). Normalised ankle joint stiffness was greater during self-selected than forced pace hopping when AL hopping on both surfaces (**p < 0.01).**

A description of the analysis of the variance of normalised knee joint stiffness during AL hopping on flat surface at a self-selected pace is included in Appendix J.8.Supplement describing the finding of negative normalised knee joint stiffness (Figure 3.4.14). No significant difference was detected for normalised knee joint stiffness values across hopping conditions (Figure 3.4.14) (Appendix J: Figure J.8). The 95%CI for normalised knee joint stiffness during AL hopping on flat surface at a self-selected pace represents the finding of a large variance (CV = 127.5%) whereby the error bars are shown to fall below zero. There were no scores for negative knee joint excursion or normalised knee joint moment during this hopping condition. The extreme scores for normalised knee joint stiffness during AL hopping on a flat surface at a self-selected pace compared to other hopping conditions are due to the small value of angular excursion mean 0.15 radians (SD) (0.07) compared to other conditions that range form mean 0.24radians (0.09) to 0.57radians (0.07), that yield relatively smaller values for knee stiffness.

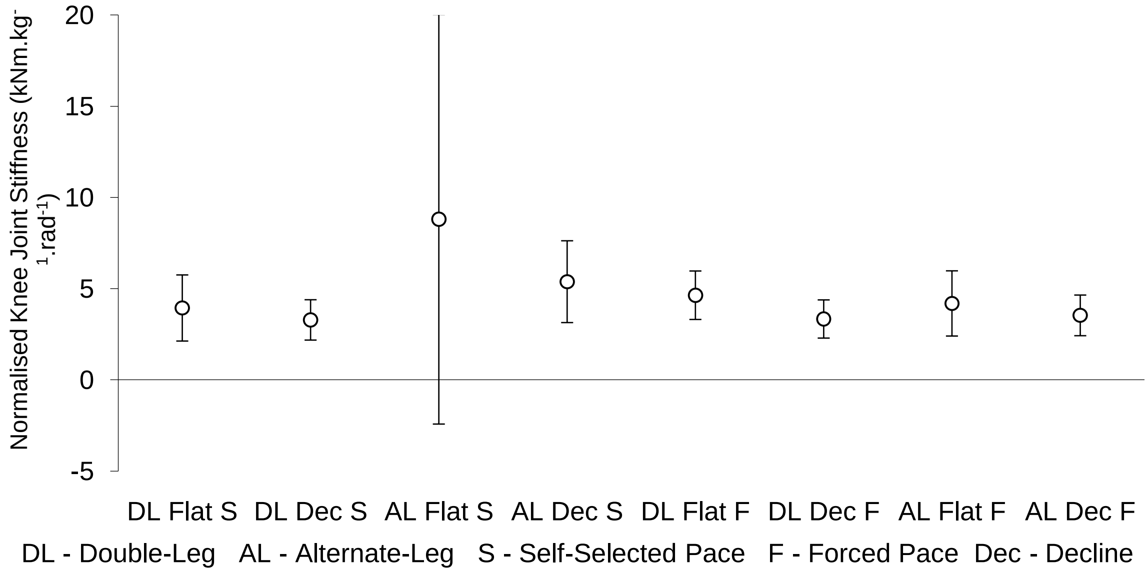


Figure 3.4.14 Normalised knee joint stiffness (kNm.kg^{-1}.rad^{-1}) (mean SD) during braking phase was not significantly between hopping conditions.

No significant difference was detected for normalised hip joint stiffness values across hopping conditions (Figure 3.4.15) (Appendix J: Figure J.9). Across all hopping conditions, 13 of 72 scores for normalised hip joint stiffness were negative. These negative scores represent movement into hip joint extension.

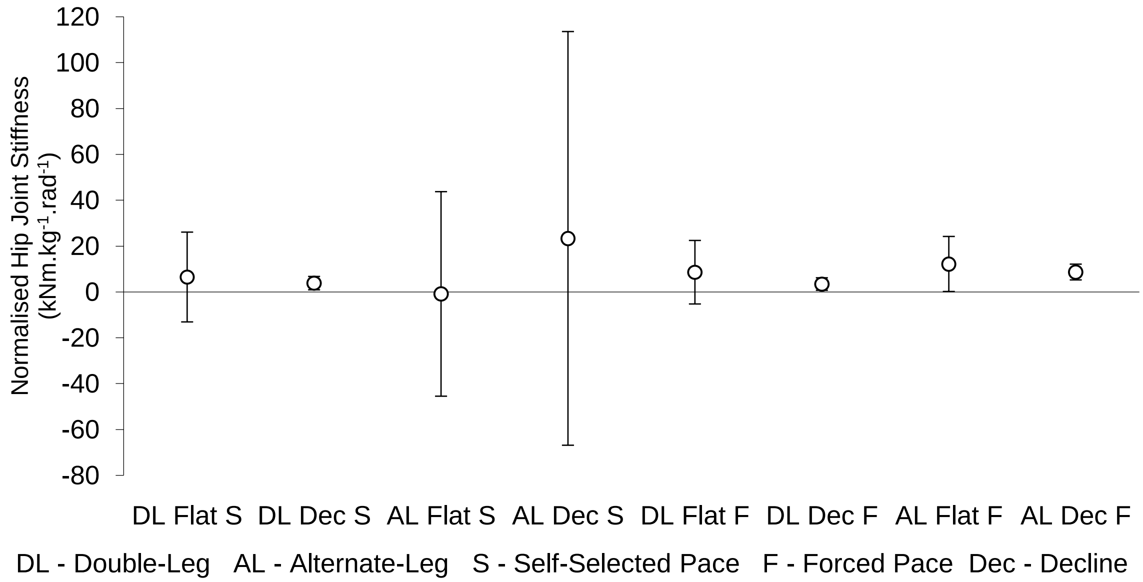


Figure 3.4.15 Normalised hip joint stiffness ($kNm.kg^{-1}.rad^{-1}$) (mean SD) during braking phase was not significantly different between hopping conditions.

The extreme scores for normalised hip joint stiffness during AL hopping on either surface at a self-selected pace are due to the small value of hip angular excursion of mean -0.002rad (SD) (0.07) and 0.05rad (0.06) respectively. These are much smaller to other hopping conditions that range from mean 0.12radians (SD) (0.06) to 0.24radians (0.07) and yield relatively large values for normalised hip joint stiffness.

3.4.2.ii – Normalised Joint Mechanical Work Done

Descriptive results are shown in Table 3.4.6 for normalised joint work at the ankle, knee and hip joints during braking phase.

Table 3.4.6 Normalised mechanical work done ($J.kg^{-1}$) (mean (SD) [%CV]) for ankle, knee and hip joints during braking phase across hopping conditions.

Hopping Condition	Ankle	Knee	Hip
DL Flat S	2.08 (0.55) [26.6]	0.51 (0.33) [65.0]	0.04 (0.09) [234.4]
DL Dec S	1.25 (0.30) [23.8]	0.79 (0.31) [39.6]]	0.14 (0.12) [87.2]
AL Flat S	2.07 (0.50) [24.3]	0.12 (0.17) [148.7]	0.20 (0.08) [41.1]
AL Dec S	1.18 (0.36) [30.7]	0.30 (0.20) [67.6]	0.19 (0.14) [74.2]
DL Flat F	2.31 (0.35) [14.9]	0.64 (0.23) [36.3]	0.08 (0.07) [88.0]
DL Dec F	1.54 (0.37) [24.0]	1.05 (0.24) [23.1]	0.13 (0.11) [80.0]
AL Flat F	2.91 (0.62) [21.3]	0.54 (0.33) [60.8]	0.32 (0.06) [18.0]
AL Dec F	1.70 (0.39) [23.0]	0.76 (0.32) [42.1]	0.35 (0.17) [49.5]

DL – Double-Leg; AL – Alternate-Leg; S – Self-Selected Pace; F – Forced Pace; Dec – Decline

Normalised ankle joint mechanical work done was significantly lower when hopping on a decline compared to flat surface for both tasks and at both paces ($p < 0.01$) (Figure 3.4.16) (Appendix J: Figure J.10). Normalised ankle joint mechanical work done was significantly greater during AL than DL hopping on a flat surface at a forced pace ($p = 0.029$) (Figure 3.4.16) (Appendix J: Figure J.10). Normalised ankle joint mechanical work done was significantly greater hopping at a forced compared to self-selected pace during DL hopping on a decline surface ($p = 0.023$), AL hopping on a flat surface ($p < 0.01$) and AL hopping on a decline surface ($p = 0.001$) (Figure 3.4.16) (Appendix J: Figure J.10).

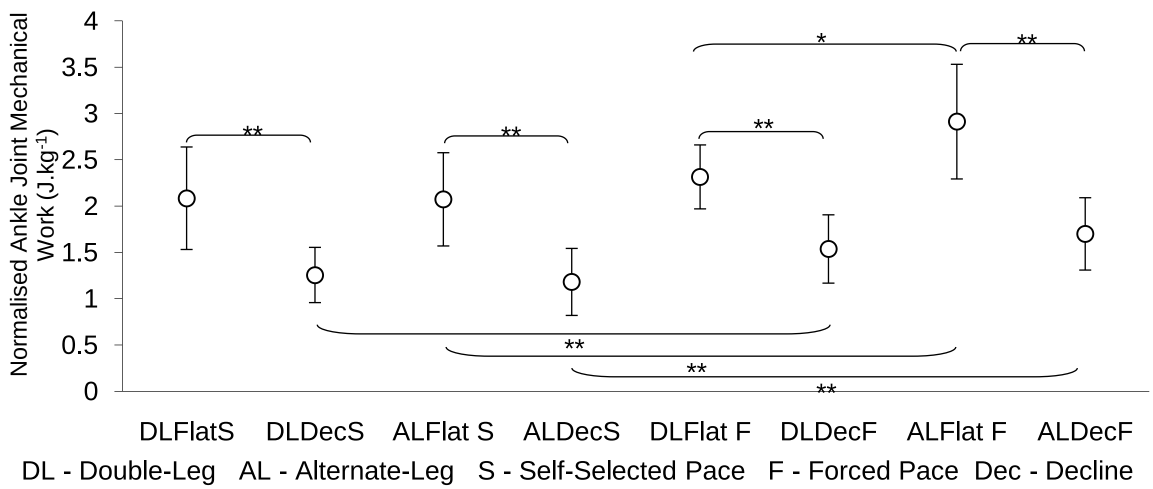


Figure 3.4.16 Normalised mechanical work done (J.kg^{-1}) (mean SD) at the ankle joint was significantly lower when hopping on a decline compared to flat surface during both tasks and at both paces ($p < 0.01$). Normalised mechanical work done at the ankle was greater during alternate-leg (AL) than double-leg (DL) hopping on a flat surface hopping at a forced (F) pace (*$p < 0.05$). Normalised mechanical work done was greater hopping at a force compared to a self-selected pace during double-leg hoping on a decline surface (**$p < 0.01$) and during AL hopping on both surfaces (**$p < 0.01$).**

Normalised knee joint mechanical work done was significantly greater hopping on a decline compared to flat surface during DL hopping at a forced pace (p = 0.001) (Figure 3.4.17) (Appendix J: Table J.14). There was a trend for greater normalised mechanical work done at the knee joint on decline compared to the flat surface during both tasks at a self-selected pace (p = 0.03) (Appendix J: Figure 3.4.17) (Table J.13). Normalised mechanical work done at the knee joint was significantly greater during DL than AL hopping on both surfaces at a self-selected pace (p < 0.01) and on decline surface at a forced pace (p = 0.01) (Figure 3.4.17) (Appendix J: Table J.13). Normalised mechanical work done at the knee joint was significantly greater during forced than self-selected pace hopping during AL hopping on both surfaces (p ≤ 0.001) (Figure 3.4.17) (Appendix J: Table J.13).

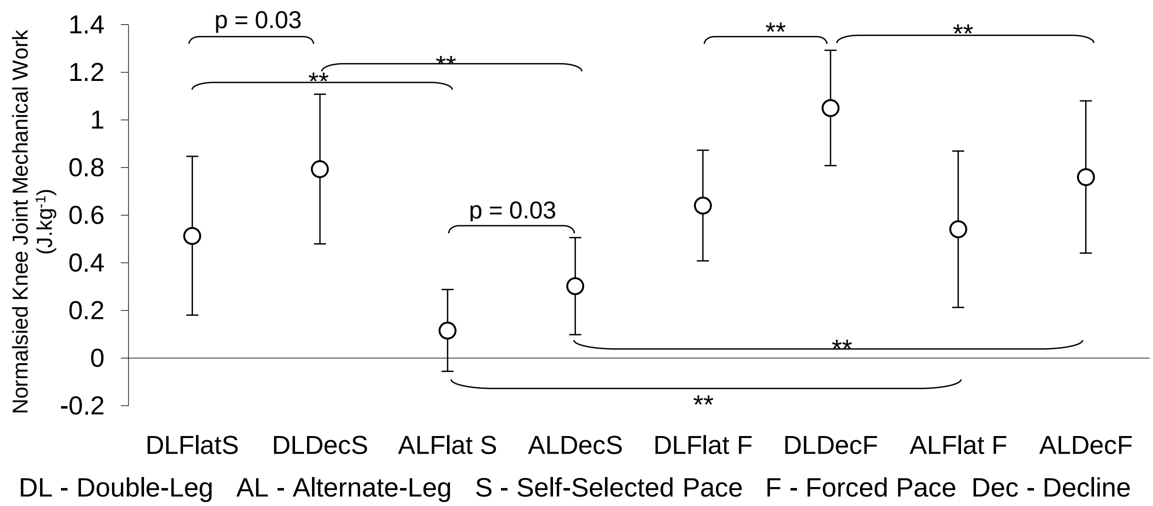


Figure 3.4.17 Normalised mechanical work done (J.kg^{-1}) (mean SD) at the knee joint was significantly greater hopping on a decline than flat surface during double-leg (DL) at a forced pace (*p < 0.05) with a similar trend at a self-selected pace (p = 0.03). Normalised mechanical work done at the knee was significantly greater during DL than alternate-leg (AL) hopping on both surfaces at a self-selected (S) pace (p < 0.01) and on decline surface at a forced pace (**p = 0.01). Normalised mechanical work done at the knee was significantly greater during forced than S pace hopping during AL hopping on both surfaces (**p < 0.01).**

Normalised hip mechanical work done was significantly greater during AL than DL hopping on a flat surface at both paces ($p < 0.01$) and on the decline surface at a forced pace ($p < 0.01$) (Figure 3.4.18) (Appendix J: Figure J.12). Normalised hip mechanical work done was significantly greater during forced than self-selected pace hopping during AL hopping on both surfaces ($p = 0.001$) (Figure 3.4.18) (Appendix J: Figure J.12).

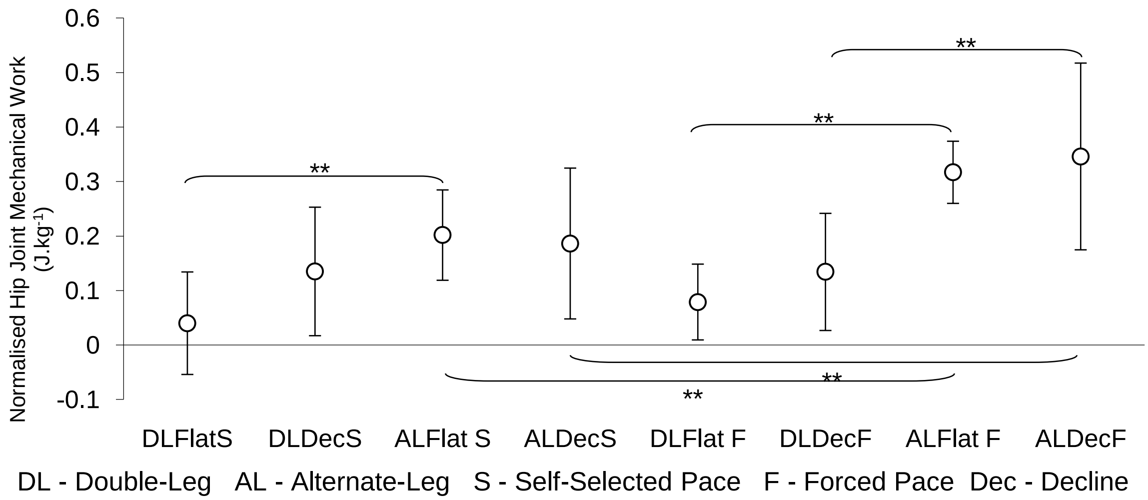


Figure 3.4.18 Normalised mechanical work done ($J.kg^{-1}$) (mean SD) at the hip joint was greater during alternate-leg (AL) than double-leg (DL) hopping on a flat surface at both paces and on a decline surface at a forced pace (*$p < 0.05$). Normalised mechanical work done was greater at a forced than self-selected pace during AL hopping on both surfaces ($p < 0.01$).**

Summary of Normalised Joint Mechanical Work Done in Braking Phase

- Normalised mechanical work done at the ankle joint was lower during decline surface hopping than flat surface hopping (Figure 3.4.16) while normalised mechanical work at the knee joint was greater on a decline surface than flat surface hopping (Figure 3.4.17).

3.4.2.iii Electromyography – Between Hopping Condition Comparisons

Statistical results for EMG derived variables including peak activation amplitude, normalised pre-loading activation amplitude and co-activation ratios for each muscle on the left and right sides across hopping conditions are in Appendix K.

3.4.2.iii - Peak Activation Amplitude

At a forced pace of hopping, the right TA (and not the left) had significantly lower peak activation amplitude during DL hopping on a flat surface than either DL hopping on a decline surface ($p = 0.049$) or AL hopping on a flat surface ($p = 0.022$) (Appendix K: Figure K.1).

There was a significant difference in peak activation amplitude of the left soleus (and not right) between flat and decline surface hopping during AL hopping at a forced pace ($p = 0.007$) (Appendix K: Figure K.2 and Table K.1).

There were no significant differences identified between hopping conditions on either left or right sides for MG (Appendix K: Figure K.3).

There were no significant differences identified between hopping conditions on either left or right sides for BF (Appendix K: Figure K.4).

The left VL peak activation amplitude was significantly lower hopping on the flat compared to the decline surface during AL hopping at a self-selected pace ($p = 0.005$) (Figure 3.4.19) (Appendix K: Figure K.5, Table K.4). An example of this difference is illustrated from recordings of vGRF and EMG signal for a single participant during AL hopping on both surfaces at a self-selected pace (Figure 3.4.20).

The same finding on the left (Figure 3.4.19) and right (Figure 3.4.21) sides for VL were detected with significantly lower peak activation amplitude during AL than DL hopping on both surfaces ($p < 0.05$) (Appendix K: Figure K.5). The same finding on the left (Figure 3.4.19) and right (Figure 3.4.21) sides for VL were detected with significantly greater peak activation amplitude during forced compared to self-selected pace of hopping ($p < 0.01$) (Appendix K: Figure K.5). An example of this is illustrated for the left VL EMG recording and vGRF for a single participant during self-selected pace hopping, during both tasks and on both surfaces (Figure 3.4.22).

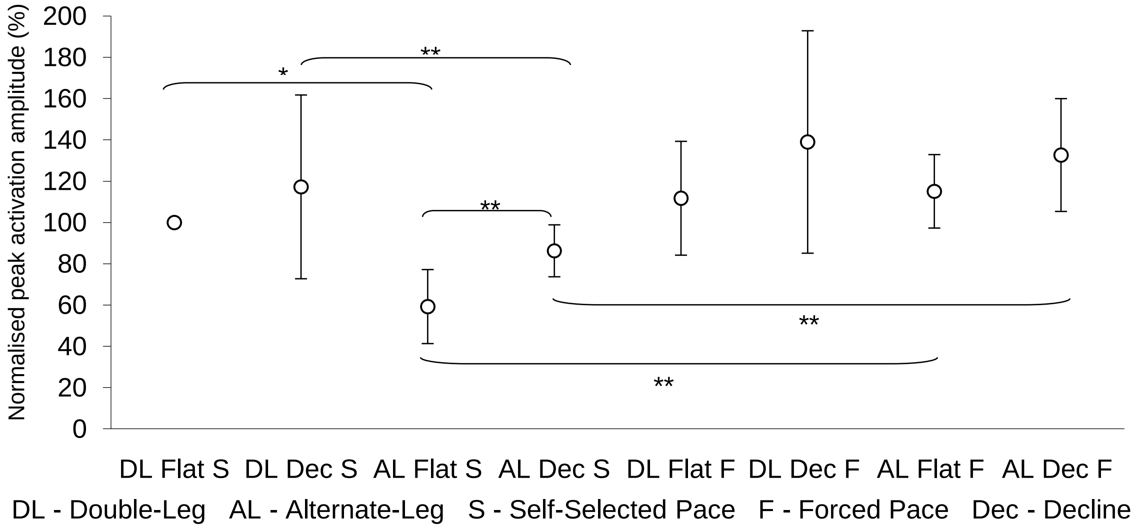


Figure 3.4.19 Left vastus lateralis (VL) peak activation amplitude (percentage normalised in figure) (mean SD) is significantly lower on the flat compared to the decline surface during alternate-leg (AL) hopping at a self-selected (S) pace (**). Left VL peak activation amplitude is significantly lower during AL than double-leg (DL) hopping on both surfaces at a self-selected hopping pace (*). Left VL peak activation amplitude is significantly greater during forced compared to S pace of hopping during AL hopping on both surfaces (**).

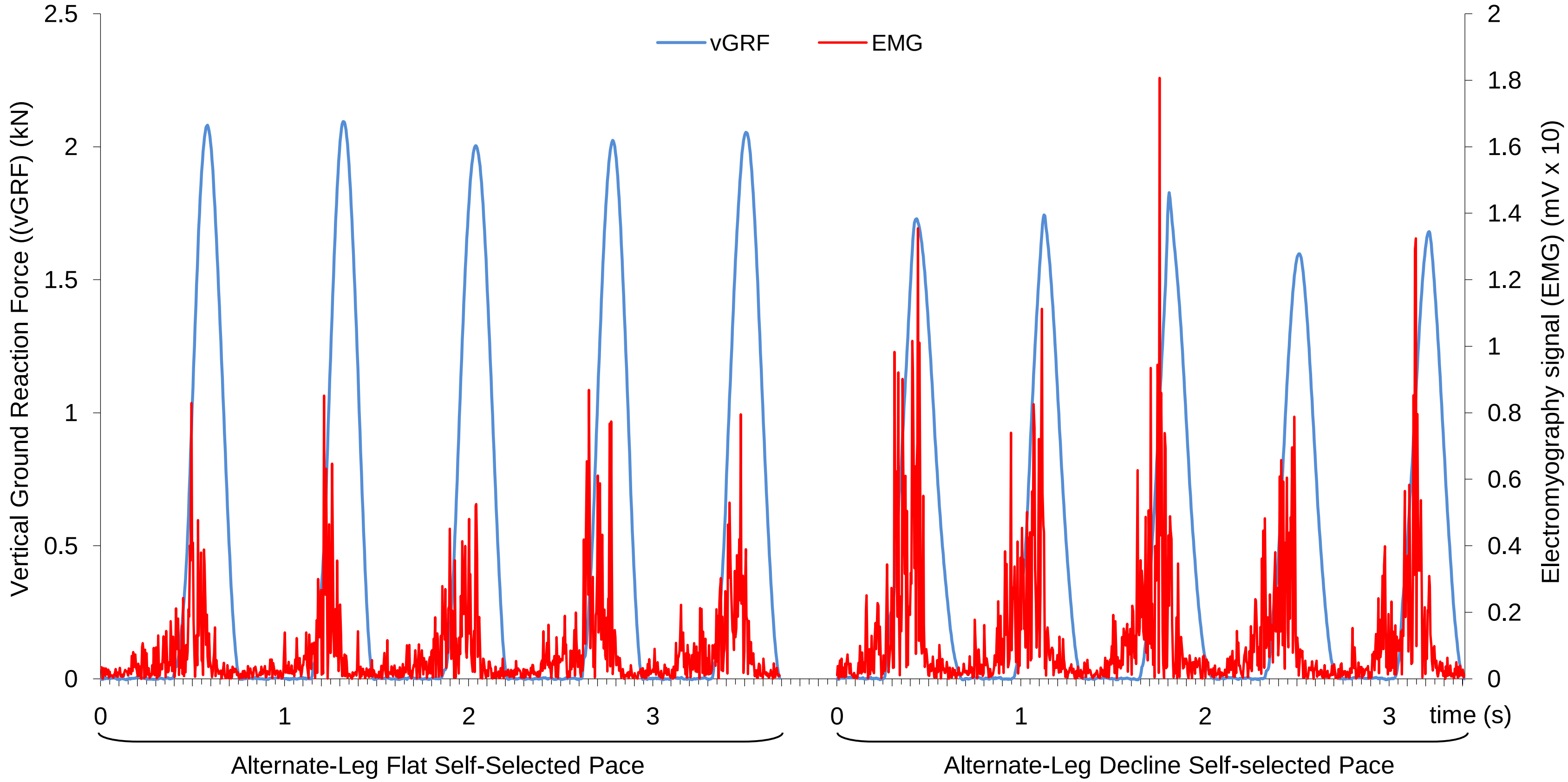


Figure 3.4.20 An example from the current study of a recording of the electromyography signal (band pass filtered from 20 to 500 Hz and rectified) of the left vastus lateralis (red line) and the vertical ground reaction force (blue line) during alternate-leg hopping on either surface at a self-selected pace, for five consecutive hops.

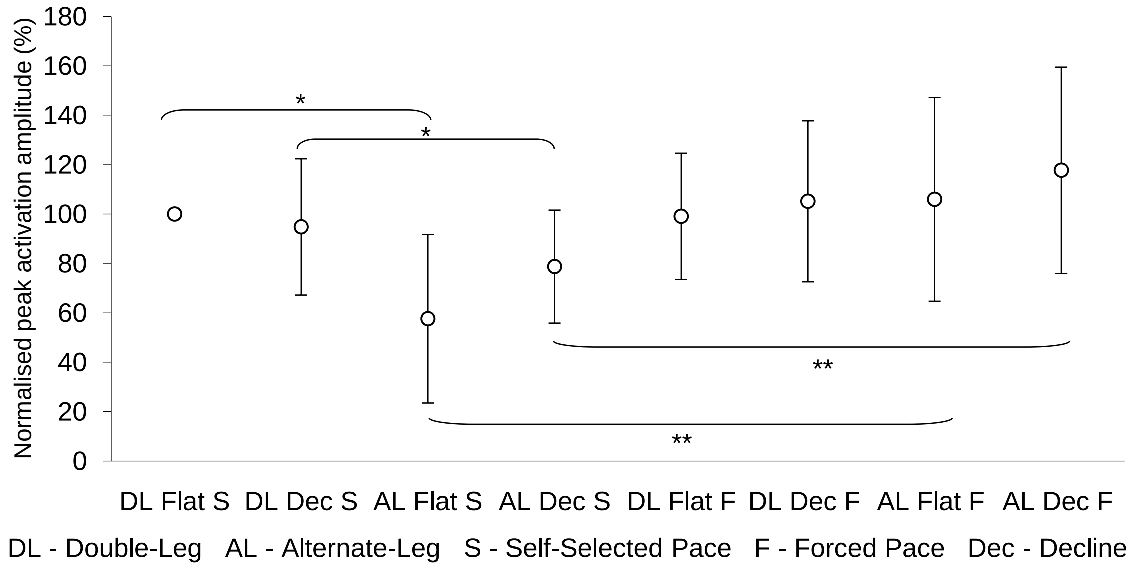


Figure 3.4.21 Right vastus lateralis (VL) peak activation amplitude (% normalised in figure) (mean SD) is significantly lower during alternate-leg (AL) than double-leg (DL) hopping on both surfaces at a self-selected (S) hopping pace (*). Right VL peak activation amplitude is significantly greater during forced compared to S pace of hopping during AL hopping on both surfaces (**).

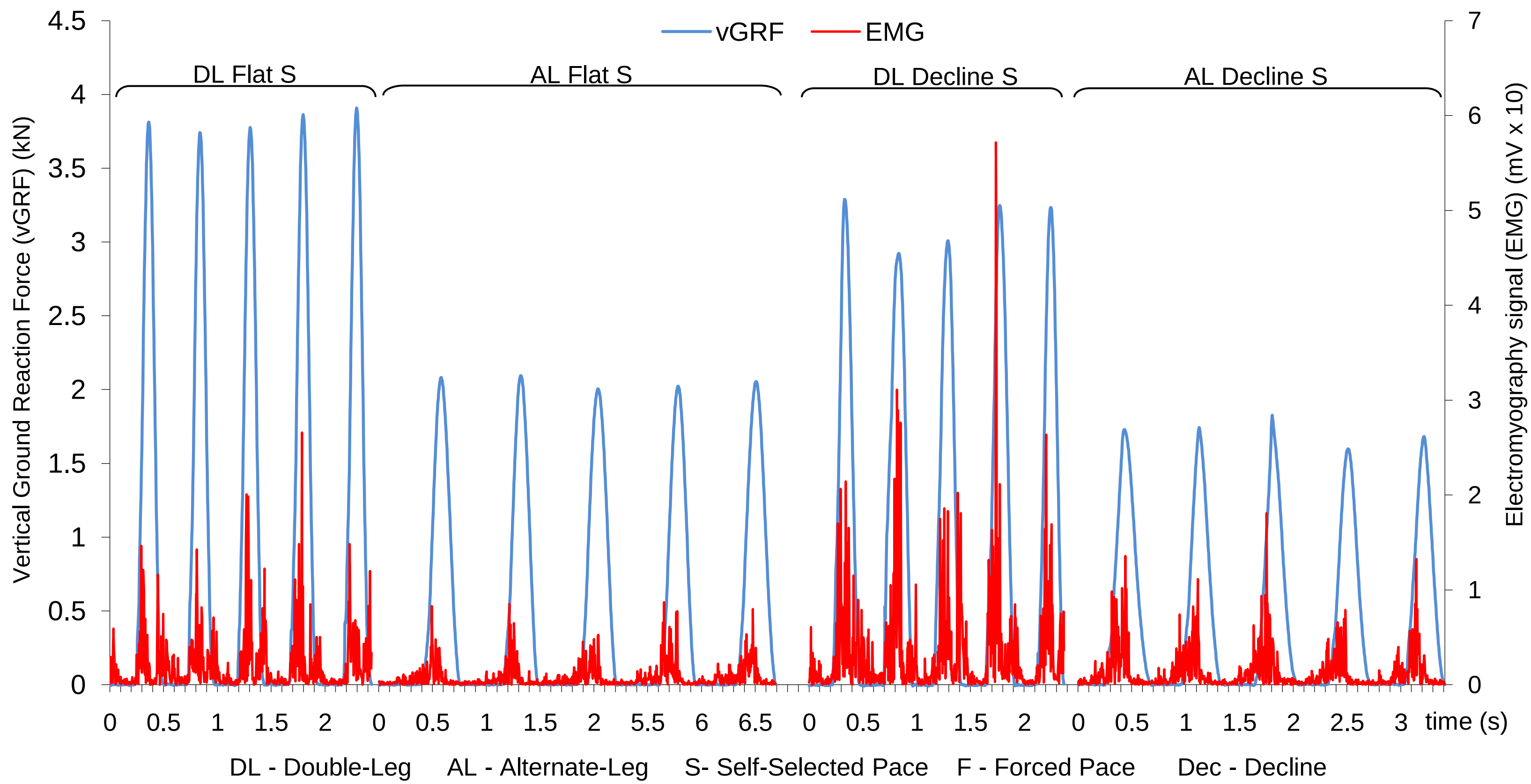


Figure 3.4.22 An example from the current study of a recording of the electromyography signal (band pass filtered from 20 to 500 Hz and rectified) of the left vastus lateralis (red line) and the vertical ground reaction force (blue line) during all hopping tasks at a self-selected pace (alternate-leg and double-leg hopping on both flat and decline surfaces) for five consecutive hops.

The left GM peak activation amplitude was significantly greater during AL than DL hopping on a decline surface at a self-selected pace (p = 0.001) and on flat surface hopping at a forced pace (p = 0.01) (Figure 3.4.23) (Appendix K: Figure K.6).

The right GM peak activation amplitude was significantly different between DL and AL (p = 0.012) (Appendix K: Figure K.6), however post hoc comparison did not detect a difference (Appendix K: Table. K.5). The lack of a significant difference may be due to a small effect size and relatively large variance. Hence, the trend of greater peak activation amplitude during AL than DL of the right GM is likely, especially when hopping on a decline surface at a self-selected pace (p = 0.02) (Appendix K: Tables K.5).

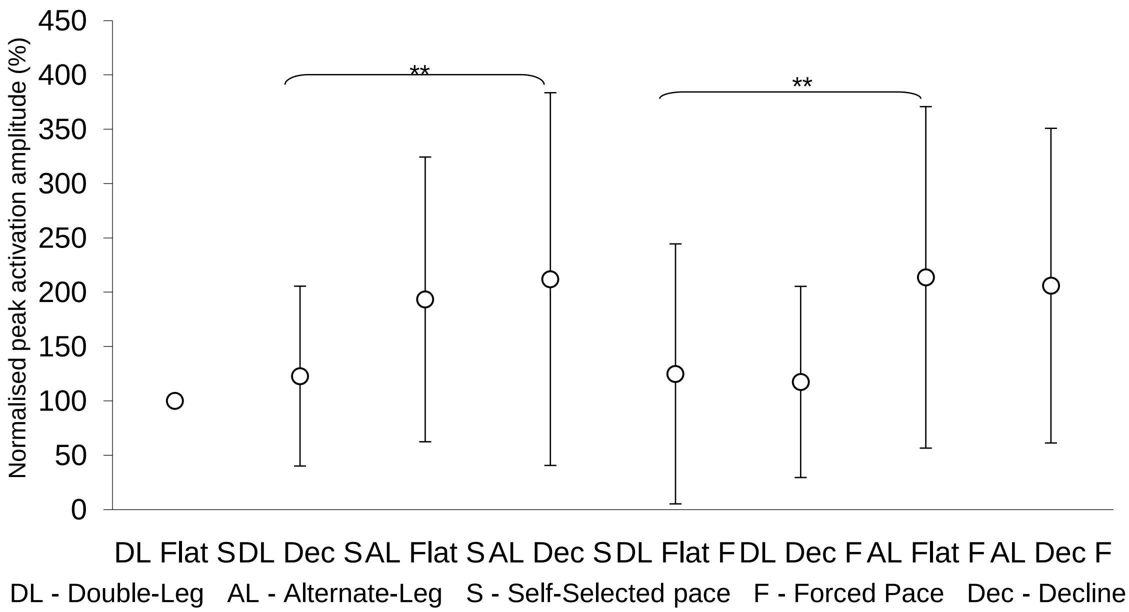


Figure 3.4.23 Left gluteus medius peak activation amplitude (% normalised in figure) (mean SD) is greater during alternate-leg than double-leg hopping on a decline surface at a self-selected pace and on a flat surface at a forced pace (**).

3.4.2.iii - Normalised Pre-Loading Activation Amplitude for medial gastrocnemius

The left MG pre-loading activation amplitude was significantly greater when hopping on the decline compared to the flat surface during DL ($p = 0.005$) and AL ($p = 0.01$) hopping at a self-selected pace and during AL hopping at a forced pace ($p = 0.005$) (Figure 3.4.24) (Appendix K: Figure K.7, Table K.6). There is a similar trend during DL hopping at a forced pace ($p = 0.07$) (Figure 3.4.24) (Appendix K: Figure K.7, Table K.6). An example of these differences are illustrated from recordings of vGRF and EMG signal for a single participant during DL hopping at a self-selected pace on flat and decline surfaces (Figure 3.4.25) and during AL hopping on both surfaces and paces (Figure 3.4.26).

The right MG has a similar trend as the left side, with greater normalised pre-loading activation amplitude when hopping on decline compared to flat surface during DL hopping at a self-selected pace ($p = 0.022$), DL hopping at a forced pace ($p = 0.07$) and during AL hopping at a forced pace ($p = 0.022$) (Appendix K: Table K.7).

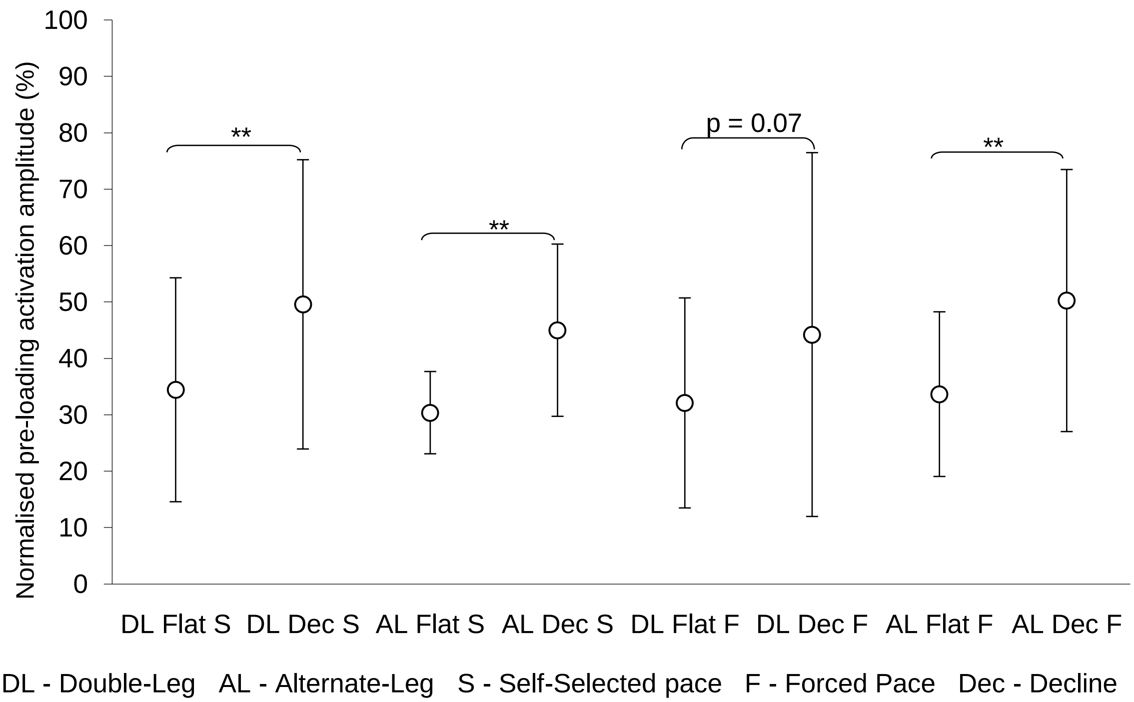


Figure 3.4.24 Left medial gastrocnemius normalised pre-loading activation amplitude (% normalised) (mean SD) is significantly greater when hopping on a decline compared to flat surface during double-leg (DL) hopping at a self selected pace (**) and alternate-leg at both paces (**). There is a similar trend during DL hopping at a forced pace (p = 0.07).

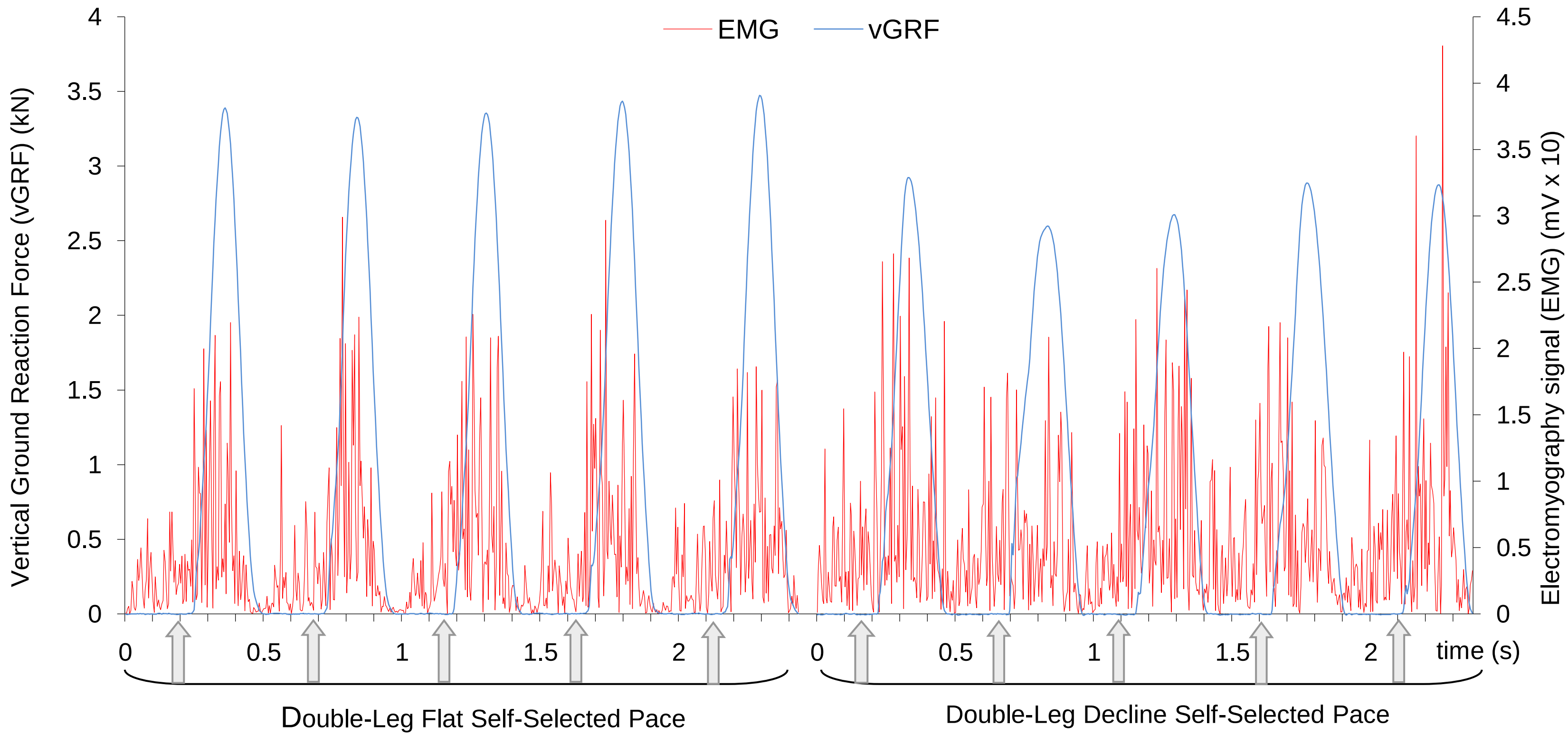


Figure 3.4.25 An example from the current study of a recording of the electromyography signal (band pass filtered from 20 to 500 Hz and rectified) of the left medial gastrocnemius (red line)and the vertical ground reaction force (blue line) during flat and decline surface hopping during double-leg hopping at a self-selected pace for five consecutive hops. Pre-loading activation amplitude was calculated for the 100ms prior to initial contact for each hop (shaded grey arrows).

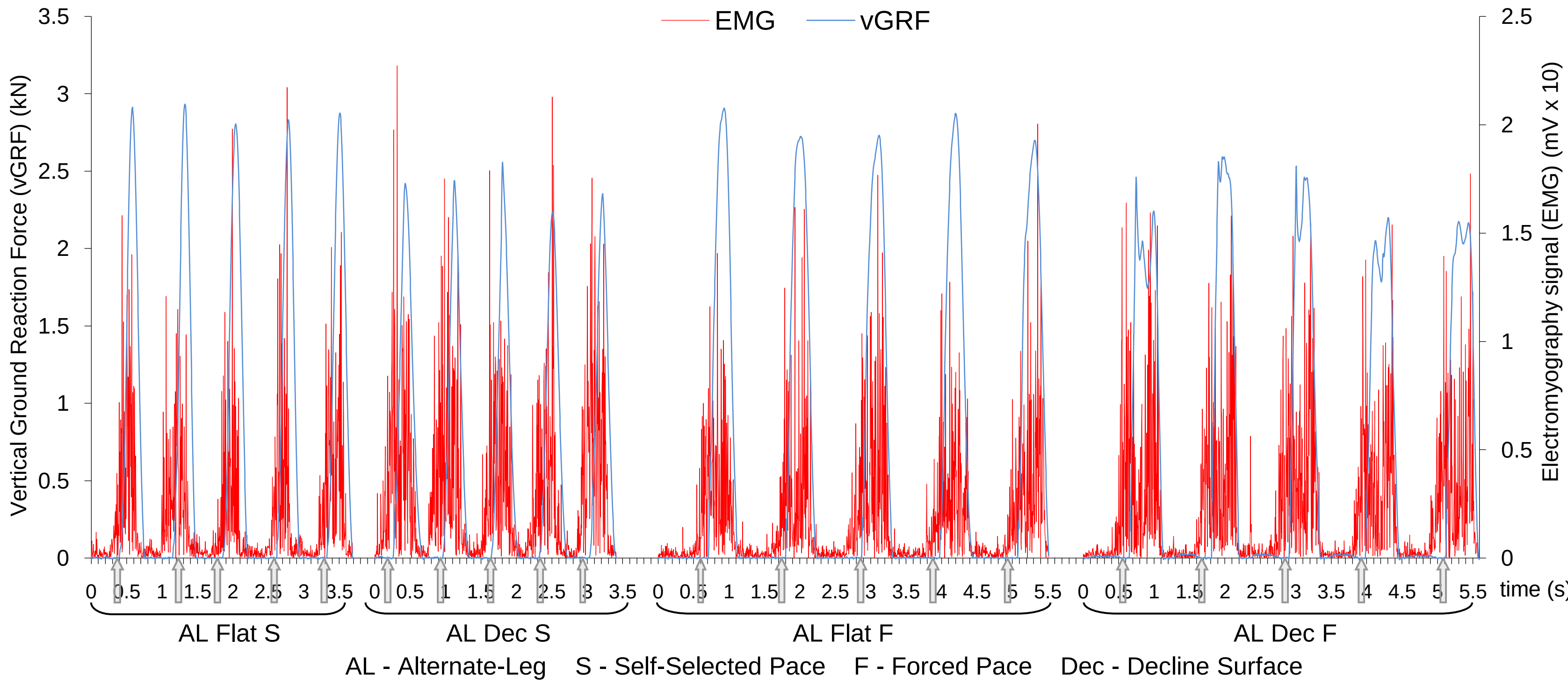


Figure 3.4.26 An example from the current study of a recording of the electromyography signal (band pass filtered from 20 to 500 Hz and rectified) of the left medial gastrocnemius (red line) and the vertical ground reaction force (blue line) during flat and decline surface hopping during alternate-leg hopping at self-selected and forced paces of hopping for five consecutive hops. Pre-loading activation amplitude was calculated for the 100ms prior to initial contact for each hop (shaded grey arrows).

3.4.2.iii - Agonist/Antagonist Co-activation Ratios

There were no significant differences across hopping conditions on the left or right for the CR between Sol and TA (CR_a) (Figure K.8), MG and TA (CR_b) (Appendix K: Figure K.9) and VL and BF (CR_c) (Appendix K: Figure K.10).

On the left there was a significantly greater co-activation ratio between VL and MG (CR_d) when hopping at a forced compared to a self-selected pace during AL hopping on a flat surface ($p = 0.013$) and AL hopping on a decline surface ($p = 0.027$) (Figure 3.4.27) (Appendix K: Figure K.11, Table K.8). There is no error bar for CR_d calculated on either the left or right sides during DL hopping on a flat surface at a self-selected pace, as this was the reference hopping condition against which all other values were normalised; hence the value of 1.

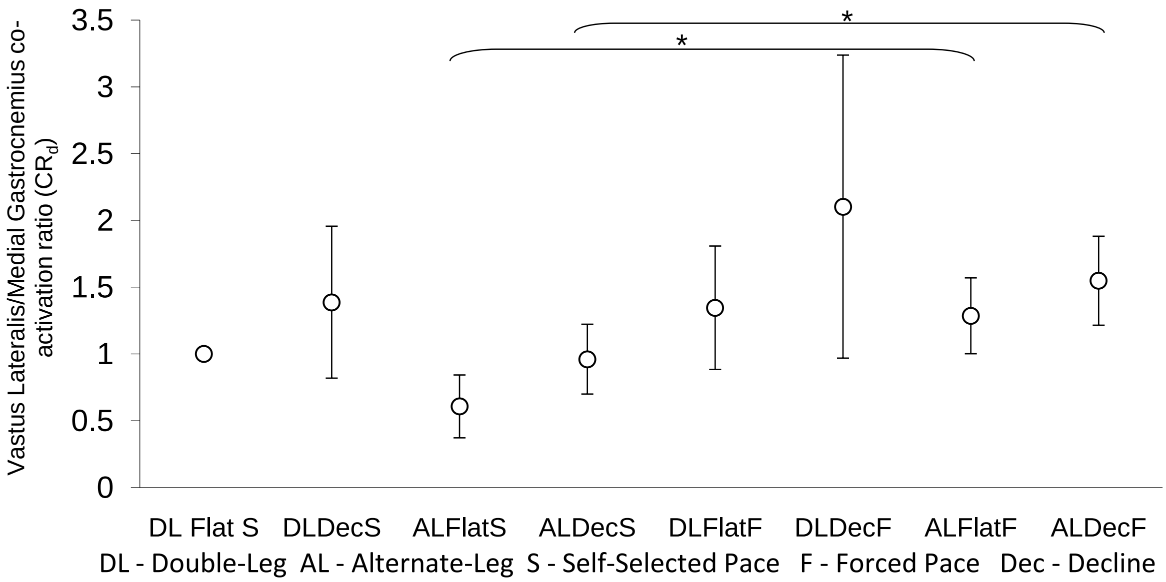
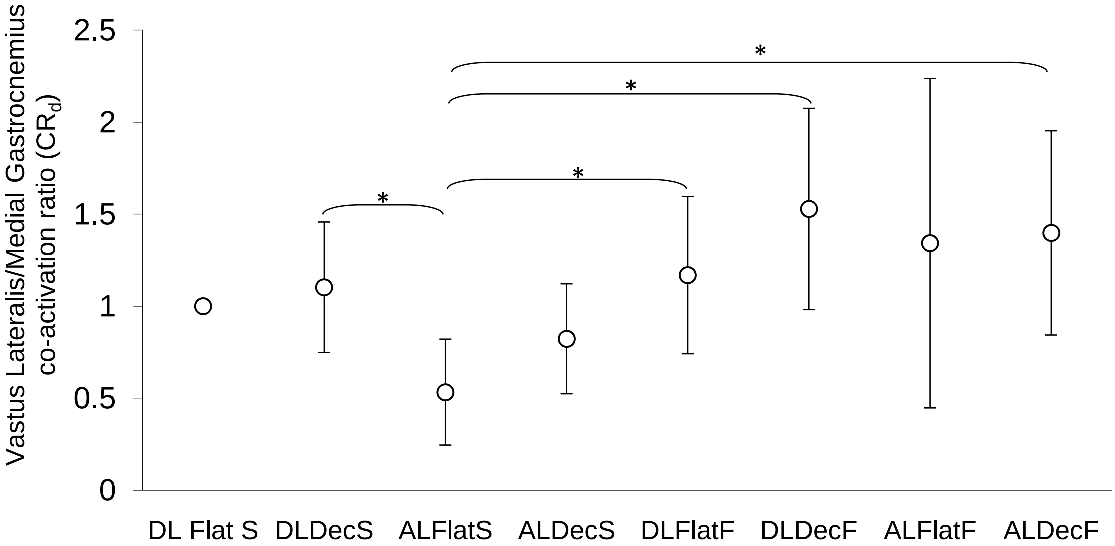


Figure 3.4.27 Left co-activation ratio (mean SD) between vastus lateralis and medial gastrocnemius was significantly greater when hopping at a forced compared to flat surface during alternate leg hopping on both surfaces (*).

On the right there was no statistically significant difference between hopping conditions for VL/MG co-activation ratio. The CR_d during AL hopping on a flat surface at a self-selected pace was significantly lower than during DL hopping on a decline surface at both paces ($p < 0.02$), DL hopping on a flat surface at a forced pace ($p = 0.04$) and AL hopping on a decline surface at a forced pace ($p = 0.001$) (Figure 3.4.28) (Appendix K: Figure K.11, Table K.9).



DL - Double-Leg AL - Alternate-Leg S - Self-Selected Pace F - Forced Pace Dec - Decline

Figure 3.4.28 Right co-activation ratio (mean SD) between vastus lateralis and medial gastrocnemius was significantly greater during forced compared to self-selected pace hopping during double-leg hopping on a decline surface (*).

Summary of findings for Co-activation Ratios

- The main findings were that there was no difference in agonist/antagonist co-activation ratios and that the finding of significantly greater CR_d (VL/MG) during the forced hopping pace was a function of greater VL peak activation amplitude at a forced pace, as there was no significant change in MG peak activation amplitude across hopping paces.

3.5 Discussion

The main finding was that at sub-maximal hopping efforts, modulation of lower limb mechanical behaviour is specific to the hopping conditions. The ankle joint is the primary site for modulation of lower limb mechanics, however, when ankle function is constrained by hopping on a decline board, knee joint mechanical behaviour adapts to maintain performance.

3.5.1 Mechanical Behaviour of the Lower Limb during Hopping

This study demonstrated that during hopping there were changes to motor control parameters at the ankle and knee joints as the level of difficulty changed across hopping conditions. There was reduced impulse as difficulty increased moving from DL to AL (single-leg) loading and from a flat to decline surface during hopping. The change in impulse was a function of reduced height and vGRF in braking phase when comparing AL to DL hopping and decline to flat surface hopping. At the sub-maximal hopping efforts tested in this study, the pattern was similar at both self-selected and forced hopping paces.

At a self-selected pace there was a clear reduction in vertical height displacement in flight during AL compared to DL hopping (on both surfaces) and explains the only significant difference in vertical leg stiffness, that was greater during AL hopping at a self-selected pace on either surface compared to all other hopping conditions. When the hopping frequency was not constrained (self-selected pace) the individual was found to reduce the height of hopping during the single-leg loading strategy (AL hopping). However, when the performance parameter was maintained with a hopping frequency of 1.8Hz, vertical leg stiffness was also maintained across DL and AL hopping tasks on both surfaces. This was particularly evident during DL hopping on a decline surface at 1.8Hz that had a very low variance for vertical leg stiffness with a CV of only 11% compared to the larger CV values across the other 7 hopping conditions that ranged from 15% to 60%. Thus, a global strategy of a conservation of vertical leg stiffness at a common hopping frequency regardless of laterality (DL or AL) and surface (flat or decline) was evident.

Vertical leg stiffness in the current study with a mean of $44.5kN.m^{-1}$ (SD) (9.5) (that is $0.49kN.m^{-1}.kg^{-1}$ (SD) (0.11) when normalised to body weight) was similar to values reported previously (Dyhre-Poulsen, et al., 1991; Hobara, et al., 2008; Hobara, et al., 2010b; Padua, et al., 2006; Padua, et al., 2005) during DL hopping at a self-selected pace on a flat surface that had used a first principles approach for calculation of vertical leg stiffness (Blickhan & Full, 1992; Cavagna, 1975; T. A. McMahon, et al., 1990; M. Voight, et al., 1995b) (Table 3.5.1). However, the values reported by Hobara et al. (2007) that used the spring-mass model (Dalleau, et al., 2004) had a lower value for vertical leg stiffness (Table 3.5.1) than the current study. The current investigation demonstrated that using a spring-mass model underestimates vertical leg stiffness and is primarily due to an overestimation of vertical height displacement when using a spring-mass model (chapter 4). This may be a reason for the differences in the measure of vertical leg stiffness in the current study to the findings of Hobara et al. (2007) (Hobara, et al., 2007) (Table 3.5.1) and highlights that under the hopping conditions tested, the task may not be well described by the spring-mass model.

Granata et al. (2002) also reported lower vertical leg stiffness in males (Table 3.5.1) (Granata, et al., 2002) than the current study and may be due to the sample population that was tested being older (mean age of 32.1 (SD) (8.1)) compared to the relatively young population used in the current study (mean age of 18.4 years (SD) (1.5)). Therefore, age may be a confounding variable and not allow a comparison between studies. Even though vertical leg stiffness values are similar in some of these studies, the hopping frequencies reported (~2.2Hz) are higher (Table 3.5.1) than the self-selected pace in this study (2Hz). Hence, the values in previous studies for vertical leg stiffness may be shown to be lower, as it has been previously demonstrated that as contact time decreased can vertical leg stiffness increased (Farley, et al., 1999; Hobara, et al., 2010b).

Dyhre-Poulsen et al. (1991) did not report leg stiffness, however, recorded the maximum peak GRF (4500N) and vertical height displacement (0.1m). This allowed an estimate of vertical leg stiffness (45kN.m^{-1}) on the participants that included men's volleyball players. This result most closely resembles the values in the current study (44.1kN.m^{-1}) during DL hopping on a flat surface and reaffirms the notion that the calculated vertical leg stiffness must be specific to the method for calculation and to the sample population. For an individual, the self-selected hopping frequency may represent the natural resonant frequency for hopping. This hopping frequency is influenced by a number of factors including the biomechanical degrees of freedom (Auyang, Yen, & Chang, 2009; Bernstein, 1967; Yen, Auyang, & Chang, 2009a), changes to joint angles (Farley, et al., 1998; Moritz & Farley, 2004a; van der Krogt, de Graaf, Farley, Moritz, Casius, & Bobbert, 2009), intrinsic properties of the muscle-tendon-units (Moritz, et al., 2004a; van Soest & Bobbert, 1993), neural feed-forward and feedback characteristics (Dickinson, Farley, Full, Koehl, Kram, & Lehman, 2000; Esposti, Cavallari, & Baldissera, 2007; Geyer, Seyfarth, & Blickhan, 2003; Hatsopoulos, 1996; Moritz, et al., 2005; Moritz, Greene, & Farley, 2004b) and capacity to store and release elastic energy (Cavagna, Heglund, & Taylor, 1977; Cavagna, Saibene, & Margaria, 1964; Dickinson, et al., 2000; Sawicki, et al., 2009). Most notably the individual will select a hopping frequency that reduces the metabolic cost of hopping similar to the changes observed in running (Cavagna, Mantovani, Willems, & Musch, 1997; Cavanagh & Williams, 1982; Kram & Taylor, 1990). For these reasons the current study determined lower limb mechanical behaviour for a homogenous group of elite level jumping athletes, aiming to reduce the effect of variance in each of these factors on the derived variables.

Austin et al. (2002) reported that vertical leg stiffness was maintained at a similar range during single-leg hopping at a preferred hopping frequency between conditions of no added mass, plus 10% body mass and plus 20% body mass (Austin, et al., 2002a) (Table 3.5.1). Even though the values for vertical leg stiffness are relatively underestimated compared to previous studies (Table 3.5.1), the findings of Austin et al. (2002) highlight that the conservation of vertical leg stiffness in a within-subject comparison that is a common feature even with significant change to the apparent mass of the individual.

As there is maintenance of vertical leg stiffness at similar hopping frequencies during both tasks and surfaces, mechanical behaviour of the joints must adapt. This involves the modulation of ankle, knee and hip joint variables that interact to maintain a ‘steady state’ for vertical leg stiffness at similar hopping frequencies.

Table 3.5.1 Results of previous studies that tested double-leg or single-leg hopping at a self-selected (preferred) hopping frequency on a flat surface and a single study that assessed landing from a block-jump for volleyball players. Values are mean (SD) other than where indicated.

Study	Method	Sample Population	Hopping Frequency (Hz)	Contact Time (s)	Flight Time (s)	Peak Reaction Force (N)	Vertical height displacement (cm)	Vertical leg Stiffness ($kN.m^{-1}$)
Current Study - DL	First principles	Trained (♂)	1.97 (0.23)	0.27 (0.05)	0.25 (0.06)	3458 (803) 37.9 (7.2)$_{BM}$	8.0 (1.7)$_{zc}$	44.1 (9.5) 0.5 (0.1)$_{BM}$
(Hobara, et al., 2010a) - DL	First principles	Untrained (♂)	2.14 (0.06)	[a]0.24 (0.04)	[a]0.23 (0.04)	[a]39.2 (7.3)$_{BM}$	[a]11.2 (1.2)$_{zt}$	~ [a]0.3$_{BM}$
		Endurance (♂)	2.21 (0.05)	[a]0.18 (0.02)	[a]0.27 (0.02)	[a]51.3 (6.8)$_{BM}$	[a]8.4 (0.9)$_{zt}$	~[a]0.5$_{BM}$
(Hobara, et al., 2008) - DL	First principles	Endurance (♂)	1.5	[a]0.21 (0.02)	[a]0.44 (0.02)	[a]59.4 (8.85)$_{BM}$	12.8 (1.8)$_{zt}$	~ [a]0.4$_{BM}$
		Power (♂)	1.5	[a]0.18 (0.13)	[a]0.48 (0.02)	[a]68.3 (6.90)$_{BM}$	12.6 (1.7)$_{zt}$	~ [a]0.5$_{BM}$
(Hobara, et al., 2007) - DL	Spring-Mass Model	Healthy (♂)	2.1 (0.09)	~0.24	~0.34			24.3 (5.6)
(Padua, et al., 2006) - DL	First principles	Healthy (♂)	Preferred[†] & 3.0					0.49 (0.14)$_{BM}$
		Healthy (♀)						0.50 (0.20)$_{BM}$
(Padua, et al., 2005) - DL	First principles	Healthy (♂)	2.3 (0.35)					0.36 (0.11)$_{BM}$
		Healthy (♀)	2.3 (0.35)					0.33 (0.15)$_{BM}$
(Granata, et al., 2002) - DL	First principles	Healthy (♂)	2.3 (0.34)	65% (8)$_{DC}$		3.0 (0.5)$_{BW}$	8.2 (2.0)	[a]26 (9)
		Healthy (♀)	2.31 (0.35)	65% (8)$_{DC}$		2.9 (0.5)$_{BW}$	8.6 (2.1)	[a]19 (8)
(Farley, et al., 1999) - DL	First principles (values are mean (SEM))	Healthy (♂)	2.2	0.31 ±0.01		1740 ±87	2.7 ±0.3$_{zf}$ 1.21 ±0.3$_{zc}$	14.5 ±0.7
(Dyhre-Poulsen, et al., 1991) - DL	First principles	Volleyball (♂)	1.9	~0.23		4500$_{max}$	~30$_{zt}$ ~10$_{zc}$	*45kN.m^{-1}$_{CE}$*
(Hughes, et al., 2008) - Block-Jump	First principles	Volleyball (♂)				[a]3.64 (1.01)$_{BW}$	[a]16.7 (4.4)	[a]15.02 (8.82)
		Volleyball (♀)				[a]2.87 (0.60)$_{BW}$	[a]16.1 (4.8)	[a]10.29 (3.56)

DL – Double-Leg hopping; Block-Jump - Simulated 'block jump' landing specific to volleyball; a – significantly different between groups; BM – normalised to Body Mass (kg);); $_{zt}$ – total vertical height displacement; $_{zf}$ – vertical height displacement in flight phase; $_{zc}$ – vertical height displacement in contact phase; max – maximum; [†] - Preferred hopping frequency not reported & data reported is pooled across both hopping frequencies; DC – Duty cycle (percentage of time spent in contact of the full hop cycle that is flight plus contact time); BW – Body Weight units (i.e.; factor of body weight); $_{CE}$ – Calculated Estimate from reported data in paper.

Table 3.5.1 (continued) Method and results (mean (SD)) of previous studies that tested double-leg or single-leg hopping at a self-selected (preferred) hopping frequency on a flat surface.

Study	Method	Sample Population	Hopping Frequency (Hz)	Contact Time (s)	Vertical Leg Stiffness ($kN.m^{-1}$)
(McLachlan, Murphy, Watsford, & Rees, 2006) - DL	First principles	Healthy (♀)	2.2_{pref}		16.39 (3.17)
			2.2_{max}		19.09 (2.61
			3.2_{pref}		29.55 (3.16)
			3.2_{max}		30.59 (3.14)
(Austin, et al., 2003) - SL	First principles	Healthy (♂)	Frequency = PF-20%	[a,c]$0.33\ (0.08)_{BM+10\%}$ [b,c]$0.34\ (0.08)_{BM+20\%}$	[a]$9.5\ (5.2)_{BM+10\%}$ [b]$7.4\ (4.5)_{BM+20\%}$
			Preferred Frequency (PF)	[a,c]$0.23\ (0.05)_{BM+10\%}$ [b,c]$0.25\ (0.05)_{BM+20\%}$	[a]$17.4\ (6.9)_{BM+10\%}$ [b]$14.9\ (5.9)_{BM+20\%}$
			Frequency = PF+20%	[a,c]$0.18\ (0.03)_{BM+10\%}$ [b,c]$0.25\ (0.05)_{BM+20\%}$	[a]$25.4\ (8.3)_{BM+10\%}$ [b]$22.9\ (7.6)_{BM+20\%}$
(Austin, et al., 2002b) - SL	First principles	Healthy (♂)	Frequency = PF-20%		[a]8.9 (6.0)
			PF		[a,b]14.6 (6.6)
			Frequency = PF+20%		[b]22.7 (6.6)
(Austin, et al., 2002a) - SL	First principles	Healthy (♂)	BM		15.9 (5.7)
			BM+10%		15.96 (5.5)
			BM+20%		15.3 (5.5)

DL – Double-Leg Hopping; SL – Single-Leg Hopping; pref – preferred height hopping; max – maximal height hopping; BM – Normalised to Body Mass (kg); BM – Body Mass; a, b, c – significantly different between hopping conditions

3.5.2 Joint Variables

<u>3.5.2.i Mechanical Work Done</u>

Across all hopping conditions mechanical work done at the ankle joint was greater than the knee or hip joints. The main finding was that hopping surface has the greatest influence on joint mechanical work done. Mechanical work done at the ankle joint was significantly lower on a decline compared to flat surface, ranging from a mean of 32% (SD) (18) to 44% (11) less for both hopping paces and tasks. This trend was reversed at the knee joint, with greater mechanical work done when hopping on decline compared to flat surface, ranging from a mean of 20% (SD) (60) to 38% (22) for both hopping paces and tasks.

The CV was similar across all hopping conditions at the ankle (15% to 31%) and had a greater range at the knee joint (23% to 149%). This demonstrates that the mechanical behaviour of the knee joint may be more heterogeneous than the ankle joint and possibly reflects that either there is greater variability of a single motor control strategy at the knee joint or that there may be more than one control strategy being utilised at the knee joint as hopping conditions change. Hopping on a decline surface reduced the ankle joint's ability to absorb mechanical work during the braking phase. At a fixed hopping frequency when hopping on a flat surface and when mechanical work done at the ankle joint is greatest, mechanical work done at the knee joint was lowest.

Mechanical work done at the ankle joint was conserved when hopping frequency was increased from 2.0Hz to 2.4Hz comparing DL to AL and increased as hopping frequency was reduced from 2.0Hz to 1.8Hz for DL hopping and from 2.4Hz to 1.8Hz for AL hopping. This highlights the role of the ankle joint as the site of modulation of lower limb mechanical behaviour.

At the hip joint, the only factor found to influence the mechanical work done was that of hopping task with AL hopping having greater values for hip joint mechanical work done than DL hopping on both surfaces. Hip joint mechanical work done during braking phase was consistently positive or in the direction of flexion with minimal hip joint excursion across all hopping conditions. Only during DL hopping on a flat surface at a self-selected pace, was the standard deviation from the mean less than zero. This was largely due to negative hip joint moments for two of the nine participants. The group mean for mechanical work done at the hip joint was low and is explained primarily by the requirement in hopping to maintain the COM over the base of support. This skill requirement avoids the generation of large hip moments as it lies near the COM in the hopping posture and has only a small lever arm compared to the knee and ankle joints.

3.5.2.ii Kinematics and Stiffness

Mechanical work done and ankle joint stiffness were lower on a decline compared to flat surface. At a fixed hopping frequency there was a strong trend for lower ankle joint stiffness on decline compared to the flat surface during DL and AL hopping tasks ($p = 0.019$) (Table J.10). It was evident that hopping on a decline surface reduced ankle joint stiffness compared to flat surface hopping and is a combination of both lower ankle joint moments and lower ankle excursion when hopping on a decline compared to a flat surface.

When hopping pace was unconstrained the ankle joint was the primary modulator of lower limb mechanical behaviour as evidenced by an increase in joint stiffness from decline to flat surfaces and from DL to AL tasks. By reducing the vertical height displacement from DL to AL, the motor control system simply stiffened the ankle joint and leg (increased vertical leg stiffness). In this scenario there was a net conservation of mechanical work at the ankle joint and reduction at the knee joint from DL to AL hopping. Ankle joint stiffness appears to be the primary mechanical characteristic that changed across factors for hopping. However, the mechanical energy required to absorb kinetic energy in braking phase saw an inverse relationship between the ankle and knee joints when the factor of surface was considered. This inverse relationship was more evident at a fixed hopping frequency during DL hopping on a decline surface and AL hopping on a flat surface. Although McLachlan et al. (2006) demonstrated a poor correlation between vertical leg stiffness (tested during hopping) and ankle joint stiffness (tested using a damped oscillation technique), this may be largely due to a difference in testing methodology (McLachlan, et al., 2006) and supports the premise that the modulation of global leg stiffness is complex.

No difference was detected across hopping conditions for either knee or hip joint stiffness. The primary reason for a conservation of knee joint stiffness (moment/angular excursion) was a significant change in knee joint angular excursion. At a forced pace (1.8Hz) there was significantly greater knee joint flexion during braking when DL hopping on a decline surface compared to DL hopping on the flat and AL hopping on a decline. Yeow et al. (2010) reported similar findings when landing from different heights, with mechanical work done during the braking phase being greater in DL than the single-leg condition as well as greater knee joint flexion with these differences more pronounced at a greater jump height (Yeow, et al., 2010). Kellis et al. (2009) reported increased knee joint flexion and reduced vGRF were following knee extensor fatigue in a drop jump (30cm) task. In the current study, hopping on a decline compared to flat surface was may be more difficult and may be due to either; a greater component of the GRF directed in the anterior-posterior direction or due to a shift in work from the ankle to the knee due to a change in the energetic cost of the contraction which may result from a difference in the joint range of motion during braking between surfaces. Thus, the individual may require greater effort to maintain the same landing position on the decline surface. The finding of a change in limb geometry at the knee appears to be a key strategy by the motor control system as landing difficulty increases when hopping on the decline compared to the flat surface.

Joint angle excursions at the ankle, knee and hip in the current study, tested at a sub-maximal effort, were similar for those reported in a simulated ‘block’ jump task used in volleyball that was a near-maximal effort (Hughes, et al., 2008). Joint excursions in the current study were within 2SD from the mean reported by Hughes et al. (2008) at the ankle and hip for all hopping tasks performed between 1.8 and 2.0Hz (i.e.; excludes AL hopping tasks at a self-selected pace) and at the knee for DL on a decline surface (both paces) and AL on a decline (forced pace) (Hughes, et al., 2008). Although the study by Hughes et al. (2008) Hughes et al. (2008) tested participants to a maximal effort compared to the sub-maximal efforts in the current study, a comparison between joint excursions may be made as both studies utilised the same motion capture and marker systems when testing volleyball players. The findings of the current study when using the novel task of DL hopping on a decline board at 1.8Hz, suggest that knee joint function may be well simulated with respect to the degree of angular displacement in the volleyball-specific task of landing from a block jump as reported by Hughes et al. (2008).

Reduced knee joint flexion angle during a simulated landing from a ‘jump shot’, specific to basketball, in male and female adolescents was found in participants with a previous history of knee pain compared to matched controls (Louw, Grimmer, & Vaughan, 2006). A history of knee pain may alter the landing strategy by reducing knee joint flexion. The current observations of an increase in knee joint excursion during AL hopping at a forced pace compared to the self-selected pace are consistent with the demand of having to land from a greater vertical height which was also significantly greater during AL hopping tasks at a forced pace. Greater knee joint flexion angles in landing from a ‘spike’ jump, specific to volleyball, were found to be predictive of patellar tendinitis in all ten cases on the left side (D. P. Richards, et al., 1996) with similarly greater knee joint flexion angle, lower knee joint excursion and concomitant hip joint extension on landing, in a stop-jump task being reported for recruits with PT abnormality compared to controls (Edwards, Steele, McGhee, Beattie, Purdam, & Cook, 2010b). In contrast, lower ankle and knee joint flexion angles in braking phase were cited as risk factors for the development of patellar tendinopathy comparing male volleyball players with a history of patellar tendinopathy to an asymptomatic group (Bisseling, et al., 2008). These studies demonstrate that greater knee joint flexion is used as a control strategy when the demand in landing increases and that an individual may avoid greater knee joint flexion angles due to a history of knee pain or that lower flexion angles may be associated with the experience of pain. Kinematic differences do not explain the differences in mechanical work done at the knee joint between DL and AL hopping, nor does it explain why a single-leg landing strategy does not follow the same pattern of simply allowing greater knee joint excursion to absorb kinetic energy in landing as a single-leg loading strategy. The current study has demonstrated that constraining ankle joint function by hopping on a decline board induces greater mechanical work done and angular excursion at the knee joint compared to the ankle joint. This was especially true during DL compared to AL loading strategies with no significant change in knee joint stiffness. This is the first study to demonstrate this interdependence of knee joint function to ankle joint function during hopping in elite level jumping athletes.

Sorenson et al. (2010) demonstrated that at the knee joint there was lower mechanical work done and no difference in angular velocity, moment or range of motion in volleyball players with a history of patellar tendinopathy compared to controls during the eccentric phase of an approach jump (Sorenson, Arya, Souza, Pollard, Salem, & Kulig, 2010). The finding by Sorenson et al. highlights the role of the PT in the storage and utilisation of energy and links tissue adaptation to knee joint loading specifically in braking phase of the SSC as no differences were found in the concentric phase (Sorenson, et al., 2010).

The pattern of modulation of knee and ankle joint stiffness may change dependant on the hopping frequency. In the current study, by assessing hopping at 1.8Hz that was only slightly less than the self-selected pace, the aim was to avoid a gross change in global strategy and to maintain the hopping skill at a sub-maximal level, thereby controlling for larger variance in performance evident during maximal or near maximal tasks (Farley, et al., 1999). In the current study, the lower hopping frequency represented a greater maximum height in flight phase, as the individual was required to contact the ground less frequently. In essence the 'effort' required to hop at a lower frequency was greater than that at a self-selected pace.

In contrast to the self-selected pace, at a forced hopping pace (1.8Hz) there was a conservation of ankle joint stiffness (moment/excursion) between DL and AL hopping tasks on both surfaces. During AL hopping, ankle joint stiffness was lower and joint excursion greater when hopping at 1.8Hz compared to 2.4Hz. This is due to the requirement of generating greater vertical height displacement in flight phase when hopping at 1.8Hz. Ankle joint moments across self-selected and forced paces were near identical on each surface. Thus, the key factor in the finding of lower ankle joint stiffness comparing hopping at 1.8Hz to 2.4Hz was greater ankle excursion in braking phase. This conserved ankle joint stiffness between AL and DL tasks at a forced pace that was otherwise different between DL and AL at a self-selected pace. Importantly, the 'cost' of conservation of ankle joint stiffness between hopping tasks was greater mechanical work done at the ankle joint during AL on the flat surface and greater mechanical work done at knee joint during DL on the decline surface. These two hopping conditions correspond with the greatest mechanical work done and joint excursions at the ankle (AL on a flat surface at a forced pace) and the knee (DL hopping on a decline surface at a forced pace) joints respectively. This study demonstrates that when ankle joint function was constrained during decline surface hopping, the increase in demand on the knee joint was greatest compared to the ankle joint especially during DL loading compared to the single-leg loading strategy (AL hopping). Furthermore, when hopping at 1.8Hz on a decline surface, the ankle joint maintained similar mechanical work done between AL and DL hopping tasks and the hip joint increased its contribution (mechanical work done) during AL hopping at 1.8Hz on a decline surface compared to DL hopping. This study clearly demonstrates that when distal joint function was constrained, there was an increase in function of the proximal joints depending on whether it was a double- or single-leg loading strategy.

A reduction in ankle joint stiffness was demonstrated comparing an ankle-foot orthosis (AFO) that resisted plantarflexion compared to an AFO that assisted plantarflexion across a range of hopping frequencies (2.2, 2.4 and 2.8Hz) (Chang, et al., 2008). Knee joint stiffness was found to be greater only when the AFO resisted plantarflexion was compared to the control AFO condition (no assistive or resistive force), across the three tested hopping frequencies. As there was no change in knee joint moment across hopping conditions, the reason for a difference in knee joint stiffness was due to knee joint excursion differences from 1º to 3º. These are considered small and may not represent clinically significant differences as the current investigation demonstrated statistically significant differences that ranged from ~4º to ~13º. Therefore, the stimulus of using an AFO device to assist or resist plantarflexion may not have been great enough to induce greater changes in knee angular excursion between hopping conditions.

Horita et al. (2002) examined drop jumps from a 50cm height and reported similar knee angular displacement as Hobara et al. (2009), who examined maximal height hopping (1.2Hz) and the reported knee joint angular displacements were 24.6º and 18.9º respectively (Hobara, et al., 2009; Horita, Komi, Nicol, & Kyröläinen, 2002). Both these studies reported similar knee joint excursion as the current study (21.8º) when knee mechanical work done was greatest as during DL hopping on a decline surface at 1.8Hz in the current study. Horita et al. (2002) demonstrated that during braking phase of a drop jump landing task there was a strong correlation between knee joint moment and both the stiffness of the knee extensor SEC and the maximum rate of force development of the knee extensors, with no significant correlation to ankle joint variables (Horita, et al., 2002). Similarly, in a repeated SSC loading task of maximal height hopping, Hobara et al. (2009) demonstrated that leg stiffness was correlated to knee joint stiffness and not ankle or hip joint stiffness (Hobara, et al., 2009).

Although a significant difference was not detected in the current study across all hopping conditions for knee joint stiffness, the current findings support the assertion that the knee plays an important role in modulating leg mechanics during braking phase at sub-maximal or near-optimal preferred hopping paces as there was an inverse relationship between the mechanical work done at the ankle and knee during brake phase between hopping on a flat and decline surface.

3.5.3 Muscle Activation Amplitude

3.5.3.i Peak Muscle Activation Amplitude

The changes in left and right VL peak activation amplitude during self-selected pace hopping tasks were reflected by the changes in knee moments. At a self-selected pace AL compared to DL had lower knee joint moments and VL peak activation amplitude. Knee joint moments and VL peak activation amplitude were greater during AL hopping tasks at a forced pace than self-selected pace on both surfaces. Greater peak activation amplitude for the VL may imply increased MTU stiffness. Zhang et al. (1998) reported increasing knee extensor activity to be associated with increased knee joint stiffness with the highest value at 30º knee flexion when tested up to 50% of MVIC (L. Q. Zhang, Nuber, Butler, Bowen, & Rymer, 1998). In the cases in the current study where systematic differences in peak activation amplitude were not detected, it is not possible to infer whether muscle force did not change. The variations in the steady state tasks assessed in the current study are fundamentally different to isometric tests at a set joint angle. Although MTU stiffness that is shown to be dependent on joint angle during a sub-maximal isometric contraction (L. Q. Zhang, et al., 1998), using peak activation amplitude as a surrogate for stiffness measures remains unclear. As observed in the current study, higher levels of muscle activation may induce greater joint moment and therefore affect muscle stiffness. It may be that muscle force and joint moment do not change linearly and that as the demand to generate and opposing torque (i.e.; knee extension) the muscle (quadriceps femoris) tries to brake during landing by increasing the level of activation. Therefore, stiffness will depend on the muscle’s ability to overcome the torque and may be observed a change in joint excursion. What is known is that during the SSC and unlike an isometric test, the contractile history (pre-stretch) of the passive structures contributes to the force development capacity of the task (Horita, et al., 2002). The relative contribution of the passive elastic component varies according to both joint angle and temporal factors. Therefore, muscle activation amplitude may not be a true reflection of

MTU stiffness that is produced as a result of contractile and non-contractile components (Lichtwark & Wilson, 2005a).

Vastus lateralis peak activation amplitude, knee joint stiffness and the knee co-activation ratio (VL/BF) were similar across hopping conditions at a forced pace in the current study. For hopping tasks at a self-selected pace changes in muscle activation may explain the mechanical characteristics and the mechanical work done at the knee. It may be proposed that maintenance of VL muscle activation amplitude allows for maintenance of knee joint moments, VL/BF co-activation ratio and knee joint stiffness during forced pace hopping conditions. The findings that EMG activation amplitudes and co-activation ratios did not significantly influence ankle, knee or hip joint stiffness are supported by a recent study (Hobara, et al., 2010a). Hobara et al. attributed the reduction in joint stiffness values to changes in leg geometry across the hopping frequencies (Hobara, et al., 2010a). As hopping frequency increased the touchdown angles at each joint decreased, reducing the moment arm length between the joint axis of rotation and the GRF (Hobara, et al., 2010a). Reducing moments thereby reduced joint stiffness values. Hobara et al. reported that as the contact time was increased with increasing hopping frequency, there was likely greater stiffness of the leg spring allowing a quick return of kinetic energy from braking to propulsion (Hobara, et al., 2010b), however, mechanical work done at each joint was not reported.

In the current study, at the ankle joint there were no significant changes detected across hopping conditions in peak activation amplitude for MG. Only the left soleus muscle had lower peak activation amplitude hopping on a decline compared to flat surface during AL hopping at a forced pace by 17.5%. Difference in the peak activation amplitude of the TA was only detected on the right side during forced pace hopping with lower values during DL hopping on the flat than both DL hopping on a decline and AL hopping on the flat. In the current study, considering the large differences in ankle joint moment (29% to 33%) and mechanical work done (34% to 43%) between hopping conditions when hopping surface was the main differentiating factor at the ankle joint, the peak activation amplitude difference found, on one side only, does not explain the large magnitude of change of mechanical characteristics detected across all hopping conditions. This may be explained by the alteration of the pickup zone of the EMG electrodes, however does not explain the differences detected in joint moments between hopping conditions. Another rationale could be that the passive structures accounted for the difference in the mechanical characteristics. In the latter case, this would be consistent with the therapeutic rationale of using the decline board for tendinopathy and may be utilised in future studies aimed at rehabilitating overuse knee injuries.

Hopping does not require a significant displacement of the COM in either the saggital or frontal planes and accordingly mechanical work done was low and joint stiffness high at the hip when compared to the knee and ankle joints. During AL hopping the single leg in contact generates a net adduction moment at the hip joint. The finding that GM has greater activation amplitude in AL compared to DL hopping tasks is expected and supported by the finding of greater EMG amplitude during single-leg compared to double-leg exercises (Krause, Jacobs, Pilger, Sather, Sibunka, & Hollman, 2009). In the current study, differences in hip joint moments in the frontal plane were not calculated as only saggital plane motion was the focus.

3.5.3.ii Co-activation Ratios

There were no differences in agonist/antagonist co-activation ratios at the ankle joint when ankle joint stiffness was significantly different across the factor of surface (at both paces) and task (at a self-selected pace). This finding is supported by previous studies that did not find a difference in co-activation ratios at the ankle joint when there were differences between hopping conditions for ankle joint stiffness comparing hopping surfaces of different stiffness (Farley, et al., 1998), preferred and short contact times at a preferred hopping frequency (Hobara, et al., 2007) and preferred to lower hopping frequencies (1.5 and 2.1Hz) (Hobara, et al., 2010a).

Pre-loading muscle activity may be a mechanism for altering co-activation with the antagonist muscles at the ankle joint (Bobbert, Yeadon, & Nigg, 1992). At the ankle the co-activation ratio in the pre-loading phase was not explored in the current study, as the modulation in the pattern of timing and amplitude of the TA was invariant to the pre-loading phase when comparing hopping surfaces (Farley, et al., 1998), contact times (Hobara, et al., 2007), endurance and power athletes (Hobara, et al., 2008), hopping frequencies (2.1 and

3.0Hz) (Hobara, et al., 2010a), genders (Padua, et al., 2005) and pre- to post fatiguing tasks (Padua, et al., 2006).

As muscle activation amplitude does not appear to have strong relationship with joint mechanical characteristics, there may be other mechanisms that modulate mechanical work done at the ankle joint. The primary mechanism proposed is modulation of stored elastic energy during the SSC (Anderson, et al., 1993; Biewener & Roberts, 2000; Ettema, 2001; Lichtwark, et al., 2005a; Magnusson, Hansen, & Kjaer, 2003b; Rassier & Herzog, 2002; Sandercock, 2005; Takarada, Hirano, Ishige, & Ishii, 1997) that describes MTU function in hopping. Muscle force output is dependent on muscle stiffness and the limb mass during rapid movements (Hof, 2003). Although it is asserted that muscle is driven by block like activation and that the amplitude of the activation is not critical (Hof, 2003), there must be muscle activation present during both eccentric and concentric phases of muscle action to maintain stiffness. The elastic nature of muscle action, especially at high velocities, will influence the force that muscle produces and therefore the stiffness of the MTU. Therefore, although there were no significant differences in peak activation amplitude for the MG or soleus muscles cross all hopping conditions and for VL across hopping conditions at 1.8 Hzit remains unclear as to what contribution of muscle force was due to the contractile and non-contractile elements.

Hobara et al. (2008) compared endurance and power trained athletes and found contrasting differences in knee and ankle joint stiffness when hopping at two different frequencies (1.5 and 3.0Hz) (Hobara, et al., 2008). Hobara et al. did not attribute differences in EMG amplitude or touchdown angle that were found to be common to both groups (Hobara, et al., 2008). The mechanism proposed is one of a difference in the ‘intrinsic’ properties of the MTU between groups, including the connective tissues (tendons, aponeurosis and ligaments) and muscle fibre composition. The current study supports the notion that the SEC contributes to mechanical work done and contributes to moments at the ankle and knee joints during the braking phase in hopping. Although there was minimal or no change in activation amplitude of the triceps surae or VL it remains unknown as to the how much of the mechanical work done was due to the contractile and non-contractile components. This finding is of importance as hopping efforts were sub-maximal and supports the concept that modulation of a SSC activity at mid-range efforts may require little increase in metabolic cost at the site of the contractile component when the non-contractiell components may be able to absorb mechanical work done. Future studies investigating rehabilitation strategies may focus on the use of DL hopping on a decline surface to increase knee extensor loading or AL hopping at a lower than preferred frequency to increase calf extensor loading. Conversely, reducing ankle and knee extensor loading, as during AL hopping compared to DL hopping at self-selected hopping frequencies may also be investigated in future studies that aim to maintain SSC muscle action.

3.5.3.iii Pre-Loading Activation Amplitude

There was a clear trend on both sides for the MG that pre-loading activation amplitude was greater when hopping on a decline compared to flat surface. For the factor of hopping surface on the left, 3 of 4 comparisons were significant and one comparison bordered significance ($p = 0.07$). On the right 2 of 4 comparisons were close to significance ($p \leq 0.02$), 1 bordered significance ($p = 0.07$) and one comparison (AL hopping at a self-selected pace) was not significant ($p = 0.3$). These differences demonstrate that there may be consistent systematic differences in pre-loading activation amplitude between flat and decline surface hopping conditions. Differences in MG pre-loading activation amplitude were previously reported comparing short to preferred contact times at a common hopping frequency (Hobara, et al., 2007), yet no difference was detected across hopping frequencies (1.5, 2.1 and 3.0Hz) (Hobara, et al., 2010a). Pre-loading activation amplitude of MG was also shown to decrease in running as obstacle height increased and the authors suggest this to be a mechanism to control the ankle angle at touch-down when greater ankle dorsiflexion was required (Müller, et al., 2010).

The role of pre-loading activation of the MG is a feed-forward or anticipatory mechanism suggested to serve as a controller at the ankle by ‘pre-tensioning’ the calf complex and therefore to resist the dorsiflexion moment at impact (Santello, 2005). The amplitude of this pre-activity appears to be related to the magnitude of the ankle moment that may be generated in landing, as pre-loading activation amplitude increased as jump height increased (Huxley, 1974; Melvill-Jones, et al., 1971; Santello & McDonagh, 1998). The timing of pre-loading activity is found to be relatively constant prior to landing ranging from 85 to 120ms

(Funase, et al., 2001; Magalhães & Goroso, 2009; Moritani, et al., 1990; Santello, et al., 1998; Thompson & McKinley, 1995). However, hopping at a maximal frequency demonstrated that pre-loading activation shifts to only 45ms prior to IC (85 and 90ms for hopping at 2.0Hz and maximal height hopping respectively) and has an amplitude that is of near equal value to braking phase for MG and soleus (Moritani, et al., 1991). Although not investigated in the current study, Moritani et al. (1991) reported there was nearly no activity in the concentric or propulsive phase, supporting the hypothesis that the utilisation of stored energy in the SEC may contribute to the force generation in the calf muscle complex in the propulsive phase. However, the lack of muscle activation signal may not necessitate that there is no activity of the contractile component that has a variable rate of relaxation and hence may still be generating force in the propulsive phase (Ettema, 1996; Hakkinen, Alen, & Komi, 1985; Shinohara, Yoshitake, Kouzaki, & Fukunaga, 2006).

Determining limb orientation prior to landing affects the moment arm length and therefore affects the magnitude of the joint moment. This was suggested as the reason for modulation of MG pre-loading in activity (Lacquaniti, 1992) including running (Müller, et al., 2010). Rather than a strategy for optimising ankle angle for IC, it was suggested that the pre-loading activation places the ankle joint in an extended position (plantarflexion) to allow the complete dorsiflexion range to be utilised in landing (Santello, 2005). In the current study, hopping required the COM to maintain the same spatial position along the transverse plane. Therefore, as the decline surface will induce a greater forward moment of inertia of the COM during contact phase in the anterior-posterior plane, a mechanism for greater force generation to resist the forward displacement could be to generate greater MG stiffness. This would allow for a greater storage of elastic energy in the braking phase that could be subsequently released during the propulsive phase by the triceps surae complex, to resist forward translatory motion of the body. Despite the lack of clarity as to the mechanism of how the muscle activity contributes to affect ankle joint mechanics, the finding of a change in MG pre-loading activation amplitude between hopping surfaces strengthens the hypothesis that an expected change in surface (decline or change in height of surface) plays a key factor.

The role of the MG in the current study appears to be elemental in its ability to adapt to the surface. The MG is also shown to be the muscle that changes its neural activity when hopping at 2.0Hz and to maximal height, with pre-loading activity commencing 150ms prior to IC and no soleus activity (Funase, et al., 2001). In a study when hopping to fatigue, it was demonstrated that the MG preferentially alters its control strategy compared to other lower leg muscles, with pre-loading activation commencing earlier in 2 of 4 recruits from a mean of 38ms (SEM) ±30 at the start to 88ms ±35 towards the end of the DL hopping task (Bonnard, Sirin, Oddsson, & Thorstensson, 1993). In the current study, MG pre-loading activation amplitude between flat and decline surfaces during DL hopping at 1.8Hz on both sides was not significant ($p = 0.07$) while all other comparisons across the factor of surface were statistically significant or close to significance ($p \leq 0.02$). These two hopping conditions (DL on flat and DL on decline both at 1.8Hz) also had similar ankle joint excursion range in braking phase (~40º). For the remaining 3 comparisons between flat and decline surfaces (DL at self-selected and AL at both paces) there was lower ankle joint excursion on the decline surface. When there is similar MG pre-loading activation amplitude there is also similar ankle joint excursion in braking. In contrast, the lower the expected ankle joint excursion in braking, the greater the MG pre-loading activation amplitude, as observed when comparing decline to flat surface hopping. This demonstrates that the MG pre-loading activation positively influences the ankle joint excursion in braking phase when the landing is expected.

The MG pre-loading activation amplitude was shown to be similar in drop landing conditions with no weight (body mass only) or added masss (+200N and +100N), while reduced mass conditions (-172N, -337N and -495N) had lower values for MG pre-loading activation amplitude (Gollhoffer, et al., 1991). Muscle activation characteristics during braking phase and push-off phase were similar across all conditions for the leg extensors (MG, soleus and vastus medialis) while the reduced mass conditions had lower values for vGRF compared to the weighted and body mass conditions (Gollhoffer, et al., 1991). These findings appear to contrast the findings of the current study that demonstrate increased MG pre-loading activation amplitude with a reduction in the vGRF (on decline compared to flat surfaces). However, the common theme is that MG pre-loading activation amplitude changes in relation to the expected stiffness that the calf complex needs to develop in preparation for impact. Gollhoffer et al. (1991) suggest that in their protocol, as MG pre-loading activation amplitude and knee and ankle joint excursion were lower in reduced loading conditions compared to the increased loading or no loading conditions, there is maintenance of tension in the MTU ready for impact (Gollhoffer, et al., 1991). In reduced weight conditions, it is likely that the calf complex will not require as a great stiffness as the loaded and body mass only conditions. In the current study, with the expectation that there was not the range of motion available for full ankle joint dorsiflexion at impact when hopping on a decline surface, there was conversely greater MG pre-loading activation amplitude compared to the flat surface, regardless of DL or AL loading strategies, that both have similar values, as do values of loaded and body mass only landings in the study by Gollhoffer et al. (1991). Therefore, MG pre-loading activation amplitude may be sensitive to the magnitude of the force at impact, however, the pre-loading activity may preferentially dictate the joint position to optimise MTU stiffness of the calf complex at impact.

In the current study, ankle joint variables were otherwise uninfluenced by EMG activation amplitudes. This suggests that MG pre-loading activation plays a role in controlling the angle of the ankle joint in an expected landing task such as hopping. Increased pre-loading activation during decline surface hopping may increase MTU stiffness and provide greater resistance to forced dorsiflexion on impact. When hopping on a decline surface the participants were not able to dorsiflex the ankle joint in an attempt to avoid heel-contact and continue hopping. When no increase in pre-loading activation amplitude was observed, as in the case of DL hopping at 1.8Hz between surfaces, there was a conservation of ankle joint excursion. Importantly, increased ankle joint stiffness was not detected as joint mechanical behaviour between hopping surfaces including mechanical work done, moment and stiffness, were consistently lower on the decline than flat surface during both tasks and at both paces. This is the first study that clearly demonstrates a relationship between MG pre-loading activation amplitude and ankle joint excursion during a repeated and rhythmical task such as hopping. A small effect size or small sample size (reduced power) may contribute to the finding of only a trend for statistical significance for MG pre-loading activation amplitude between surfaces during DL hopping at a fixed frequency of 1.8Hz ($p = 0.07$) that was statistically significant when DL hopping at 2.0Hz.

3.5.4 Knee and Ankle Interplay

The focus of this investigation was on the mechanical behaviour of the knee joint and its interplay with the ankle joint during hopping. The key findings were that the ankle joint is the more adaptable in changing its mechanical characteristics in relation to the magnitude of work that the joint performs. However, these changes were not clearly attributable to changes in muscle activation. The role of the ankle joint as the primary site for change in mechanical behaviour was evident in single-leg hopping on a flat surface when ankle, knee and hip joints altered their angular excursion due to a change in hopping frequency with only the ankle joint adjusting angular excursion due to the factor of mass (added mass of 10% or 20% of body mass) (Austin, et al., 2003). The ankle joint was found to be the most sensitive to a change in force as hopping frequency increased (2.2, 2.8 and 3.2Hz) and the variance in ankle joint torque decreased more than at the knee and hip joints (Yen & Chang, 2009b). Similarly, using optic fiber insertions of the tendon detected greater Achilles loading compared to the PT during hopping with reduced knee joint angular excursion with mean of 23º (SD) (10), while this was reversed in conditions when hopping with greater knee angular displacement of 56º (13) (Finni, et al., 2001). The current findings support the concept that the SEC of the eccentrically acting ankle extensors may contribute significantly to the mechanical work done in braking phase of repeated impact landings (Farley, et al., 1999).

A significant finding of the current study is that when hopping was at a self-selected pace there was an inverse relationship between ankle and knee joint moments and a net conservation of knee joint stiffness. When the demand of hopping was increased (greater vertical height displacement) there was a net conservation of knee joint stiffness and greater mechanical work done at the knee joint that was not explained by changes in the level of muscle activation in the braking phase. To this end, the SEC and parallel elastic component of the QF may play a significant role in storage and utilisation of elastic energy during hopping when ankle joint function was reduced as on the decline board. This will have implications on the loading of the PT in both functional activities and rehabilitation protocols.

In contrast to the sub-maximal hopping conditions imposed on participants in the current study, a previous investigation instructed the well-trained male athletes to ‘hop as high as possible’ (Hobara, et al., 2009) and demonstrated that only knee joint stiffness was correlated with the measure of leg stiffness, with a partial regression co-efficient of 0.64 compared to that of the ankle (0.37) and hip (-0.06) joints. Continuous hopping at 2.0Hz to fatigue, induced greater knee joint flexion at IC as the task progressed with VL activity also increasing as the task progressed (Bonnard, et al., 1994). Similar to the results during hopping of Bonnard et al. (1994), the findings of increased knee joint flexion and VL activity following a fatigue protocol, were reported when using a single forward hop test (Orishimo & Kremenic, 2006). In the current study there was no significant difference in knee joint stiffness across hopping conditions, however, when vertical height displacement was greatest (DL hopping on a flat surface at 1.8 Hz) knee joint stiffness was also greatest. Hence, joint mechanical behaviour could appear to be task-specific, with sub-maximal hopping efforts modulated by a change in mechanics at the ankle joint and maximal hopping efforts modulated by a change in mechanics at the knee joint.

The finding of the current study adds to the evidence that both the ankle and knee joints were involved in the modulation of lower limb mechanical adaptive strategies rather than a global strategy of one joint only. Chang et al. (2008) demonstrated that during a single-leg hopping task there was a reduction in ankle joint stiffness using an AFO device that assisted ankle joint plantarflexion, while there were increases to both knee and ankle joint stiffness when using an AFO device that resisted plantarflexion (Chang, et al., 2008). This may demonstrate that using an AFO resistive device constrains ankle joint function and that the knee joint increases its ability to modulate leg stiffness that was equal between the two hopping conditions (Chang, et al., 2008). The knee-ankle interplay was also evident during a step-off task, demonstrated by greater mechanical work done at the ankle compared to the knee and hip joints when landing with a stiff ($<90^{o}$ knee flexion) compared to soft ($>90^{o}$ knee flexion) landing technique (Devita & Skelly, 1992). Previous findings of hopping at a preferred frequency and maximal height hopping support an ankle joint modulating strategy (Farley, et al., 1999) while constraining hopping frequency above and below the preferred frequency (1.5 and 3.0Hz) and supporting a knee and hip joint modulating strategy (Hobara, et al., 2010a). Furthermore, Arya et al. (2006) compared single-leg hopping mechanical behaviour between normal and Achilles tendinopathy affected groups and detected a modulation of knee joint stiffness (Arya, Solnik, & Kulig, 2006). This may indicate that Achilles tendinopathy acted to constrain ankle joint function such that knee joint mechanical behaviour was modulated during hopping. This is similar to the current findings that utilised a sub-maximal hopping frequency and demonstrated that reducing ankle joint function by hopping on a decline board resulted in modulation of knee joint mechanical work done.

Similarly, reduced passive ankle joint dorsiflexion range of motion was associated with the presence of patellar tendinopathy in volleyball players (Malliaris, et al., 2006) and lower active dorsiflexion was demonstrated in the horizontal compared to the vertical component of a stop-jump task (Edwards, Steele, Cook, R., McGhee, & Munro, 2010a). These findings highlight the symbiotic adaptive strategy between the knee and ankle joints that are specific to the functional demands of the tasks and may predicate dysfunction at the more proximal knee joint if there is reduced energy absorbing function at the ankle joint.

The suggested reason for a change in knee joint kinematics is one that is elemental in generating joint force whereby joint moment is proportional to lever arm length. At the ankle joint, the lever arm length (the perpendicular distance between the line of the GRF and the joint axis of rotation) operates close to its maximal length in contact phase with the ankle joint angle, at IC, ranging from 10º to 30º plantarflexion. For there to be a significant increase in contribution to ankle joint moment due to increased lever arm length is near impossible with the lever (foot) only needing to be perpendicular to the GRF (0º) to provide the best case scenario (Figure 3.5.1). The lever at the ankle joint is structurally constrained as it operates near maximal length in the braking phase of hopping.

At the knee joint however, if there is a requirement of greater knee joint moment due to an increased demand in landing, even at equal muscle force, the knee joint moment may increase significantly by increasing moment arm length (Figure 3.5.1). The relatively longer length of the tibia and greater range of joint excursion at the knee allows enough freedom of motion to allow this change in strategy. Greater knee joint flexion also approaches the optimal angle for generation of knee joint moment of 70º (Ichinose, Kawakami, Ito, & Fukunaga, 1997; S. J. Pearson & Onambélé, 2005). Muscle pennation (Ettema & Huijing, 1994) and sarcomere length (Burkholder, et al., 2001) will affect the MTU length-tension relationship and therefore the joint moment. However, from a tenet of mechanical advantage, mechanical work absorbed is able to be increased with little or no increase in metabolic activity of the dynamically acting muscles during high speed movements by increasing the lever arm length, allowing the storage of elastic energy. Importantly, the potential for changes at either or both joints are task-specific. This was demonstrated by the finding of lower knee joint moment and PT force coupled with greater ankle joint excursion, in the vertical compared to horizontal landing component of a stop-jump task, with similar knee and hip joint excursions in both landing components (Edwards, et al., 2010a).

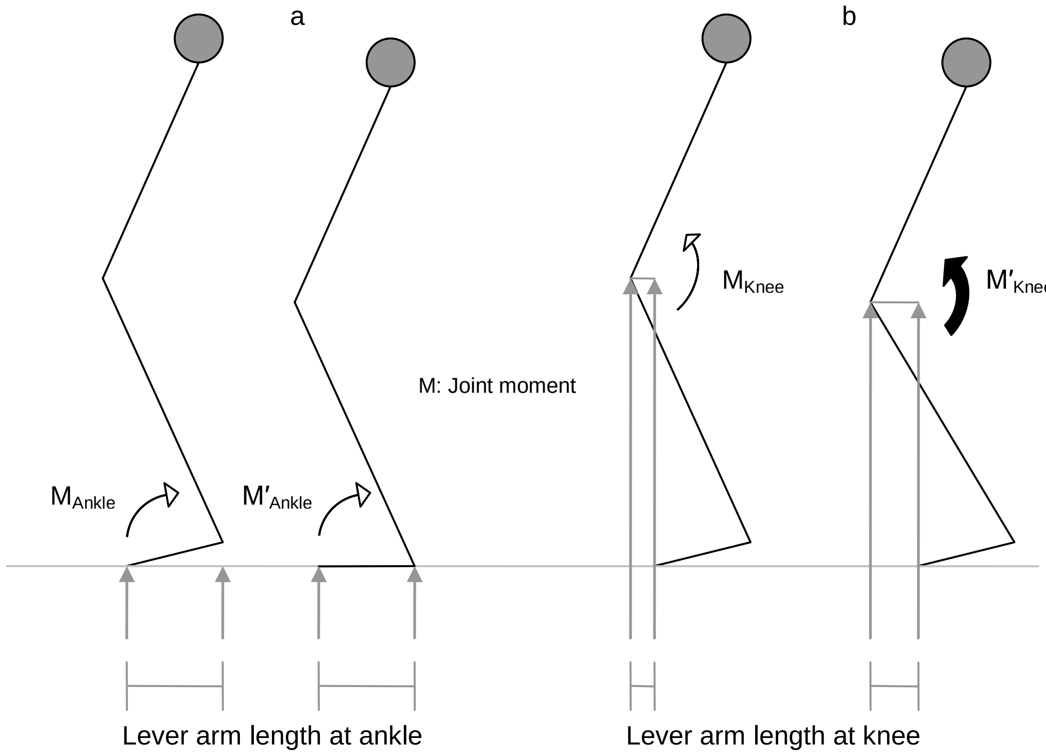


Figure 3.5.1 Schematic representation of the effect of changing ankle joint excursion (a) and the minimal change in lever arm length compared to the change in knee joint excursion (b) and relatively greater change in lever arm length. The net effect is minimal change in ankle joint moment (M_{Ankle} and M'_{Ankle}) and larger change in knee joint moment (M_{Knee} and M'_{Knee}).

Human bipedalism is a an evolutionary trait primarily amongst primates and a possible selective trait to improve feeding (Hunt, 1994) and increased capacity to carry using hands (Watson, Payne, Chamberlain, Jones, & Sellers, 2009). It remains unclear whether a significant mechanical cost advantage is achieved in bipedal versus quadrupedal gait (Niemitz, 2010; Taylor & Rowntree, 1973; Watson, et al., 2009) however, the alternating or ‘antiphase’ pattern of leg motion is described as the most common and economical at different speeds (Alexander, 1992; Peck & Turvey, 1997). The individual may voluntarily change the coordination between limbs using either an alternating-leg strategy (antiphase), double-leg strategy (jumping or hopping), galloping, unipedal hopping (Austin, et al., 2002b) and even more complex coordination patterns such as during dance (Peck, et al., 1997). The current study and previous studies on hopping and running demonstrate that the ankle joint mechanical behaviour will allow a change in motor strategy with a clear interaction with the knee joint (Farley, et al., 1998; Farley, et al., 1999; Hobara, et al., 2010a; Hobara, et al., 2007; Hobara, et al., 2009; Horita, et al., 2002) and hip (Lewis & Ferris, 2008). The common mode of using an alternating inter-limb coordination pattern (single-leg loading strategy) and an ankle joint strategy of relatively greater ankle joint stiffness during AL hopping supports the use of AL hopping on the flat as the reference condition for normalisation of muscle activation amplitude variables. This study, however, was consistent with the majority of studies that tested DL hopping and have used this strategy for normalisation for EMG derived variables (Farley, et al., 1998; Funase, et al., 2001; Hobara, et al., 2007; Moritz, et al., 2005; Müller, et al., 2010; Padua, et al., 2005).

There was no inter-limb asymmetry evident in the derived variables assessed in the current study. The lack of inter-side differences in kinetics and kinematics during hopping in the current study may be due to a training effect whereby inter-side differences are reduced over time with exposure to relatively equal volumes of loading for each leg. Inter-side differences may exist, however, are not sensitive to testing using hopping as a test and may be more evident in task-specific testing as utilised by Teixeira et al. (2003) (Teixeira, Silva, & Carvalho, 2003). There are also inherent differences in coordination patterns between the sport of soccer and jumping sports (volleyball and basketball) with the former requiring greater foot-eye coordination and the latter requiring greater hand-eye coordination. The issue of laterality is one that is complex and requires further investigation in future studies.

3.5.5 Strengths and Limitations

An important benefit is the use of a repeated measures study design allowing a within-subject analysis of changes across different hopping conditions, limiting confounding variables such as differences in anthropometry and training history between participants. Kinematic markers and sEMG electrodes were maintained in place throughout the testing procedure reducing systematic error across trials. Gait events determined in the current study used the gold standard method by detecting the vGRF using a kinetic (force plate) measure identifying IC and toe-off compared to other methods that have use kinematic measures or marker trajectories to detect gait events (Ranavolo, Don, Cacchio, Serrao, Paoloni, Mangone, & Santill, 2008).

Performing testing trials at a self-selected pace before those at a forced pace minimised a learning effect to cross over between hopping at the two paces. The use of an auditory stimulus can influence mechanical behaviour for tasks that are rhythmical and can reduce the synchronistion error (Fulop, Kirby, & Coates, 1992; Lai, Shea, Bruechert, & Little, 2002; Wuyts & Buekers, 1995). Randomisation for the factors of task and surface reduced the possibility of selective bias. Allowing adequate rest periods and using a novel single-leg loading strategy of AL hopping reduced the effect of muscle fatigue that has been shown to alter joint kinematics, minimising the introduction of fatigue as a co-variant. Sampling a homogenous group of elite level jumping athletes that train on a full-time basis, as live-in athletes, at an elite level training facility limited factors other than training to influence motor control in the novel skill of hopping.

This study did not assume the vGRF to be modeled by a spring-mass model or for there to be a linear relationship between displacement and force and used a first principles approach for calculation of vertical height displacement (flight and contact) and vertical leg stiffness with direct measure of GRF using a gold standard technique (Ranavolo, et al., 2008). The comparison between reference and field test methods made in chapter 4, clearly demonstrated that the spring-mass-model did not represent the mechanical behaviour during human hopping during AL hopping or when hopping on decline surfaces and that the spring-mass model did represent mechanical behaviour of human hopping during DL hopping on a flat surface (1.8 and 2.0Hz) and on a decline surface (1.8Hz). However, the field test method (Dalleau, et al., 2004) did underestimate values of vertical leg stiffness.

A unique benefit in the current study was the use of dual force plates to record and calculate kinematic and kinetic variables for each leg independently. Although findings generally indicated symmetry in these measures including those for muscle actions, this method does not detract from other protocols that may aim to assess inter-side differences that may be present at maximal or near-maximal efforts and may be hypothesised to generate more asymmetrical loading patterns.

This study aimed to correct sources of error in calculation of knee joint moment due to differences in the recorded centre of pressure when landing on a decline board. The method used has not previously been reported. The procedure used to correct for possible differences in the centre of pressure that may exist when there is a secondary surface placed on the force plate were appropriate for hopping when there was minimal translatory motion in the x and y axes. The method used was to correct the height of the force plate by detecting IC and using the toe-marker as an estimate of the height above the ground for the participant. It would be in the interest for future studies using a similar procedure, to validate the use of the correcting procedure of using the mean corrected height of a series of hops against the corrected height for each hop when landing is not directly on the force plate.

The behaviour of the joints is constrained by physiological parameters including the joints' of range of motion, its ability to absorb force during braking and to propel the body into the subsequent flight phase. The resistance to joint motion, or stiffness, is dictated primarily by the muscles (contractile and non-contractile components) at the joint (Farley, et al., 1998; Winter, 1990). The extensor MTU at each joint provides the primary resistance in braking, controlling the absorption of kinetic energy of the falling body. Joint stiffness was calculated rather than MTU stiffness as accurate moment arm measures for individuals vary with no common gold standard measure available. However, this does not preclude the use of joint stiffness measure in a repeated measures design where there is no gross change in muscle moment arms across testing conditions. It is appreciated that there is some change in muscle moment (joint angle dependent) as the radius of curvature changes and varies throughout the joint range of motion (P. N. Smith, Refshauge, & Scarvell, 2003). Hence, joint moments provide an accurate reflection of a change in MTU force contribution and the derived variable of joint stiffness has been used extensively in similar biomechanical studies (methods section 2.2, joint variables). It is acknowledged that a limitation of the study is the assumption that the joints are maximally displaced at peak vGRF under all hopping conditions, even though this assumption is consistent with previous studies that at sub-maximal hopping efforts the ankle and knee act in synergy and almost simultaneously in braking phase (Auyang, et al., 2009; Blickhan, Seyfarth, Geyer, Grimmer, Wagner, & Gunther, 2007; Bobbert & Casius, 2011; Farley, et al., 1991; Farley, et al., 1998; Farley, et al., 1999; Yen, et al., 2009a; Yen, et al., 2009b). Future studies may aim to determine the joint angle at impact as this may influence joint loading during hopping.

A limitation in this study was the small sample size of only nine participants. Thus, it may lack external validity. Participants consisted of elite level (national level) jumping athletes in volleyball and basketball that trained full-time (~30 hours/week) over the course of a calendar year. It has been demonstrated that the motor control strategy used in landing may be specific to the type of sport that an individual participates in (Laffaye, et al., 2005); however, it may also be specific to the competitive level of each participant within each sport. In the current study participants were exposed to a number of modes of training in addition to the activities of daily living (ADL) including leisure and recreational activities. It was impossible to control the loading history of the leg(s) of recruits; however this study sampled a relatively homogenous group of elite level jumping athletes where these parameters were relatively controlled as athletes that lived in a common residence at the training facility.

Future studies may focus on examining the within-subject variance that may be of specific interest in evaluating the motor control strategy or strategies used when hopping conditions change. For example hopping frequency, controlled by an auditory stimulus, may result in a reduction in 'natural variability' when moving from an 'open skill' (self-selected pace) to a 'closed skill' (focused attention on matching contact phase with the audible tone) (Ehrlenspiel, 2001). As joint moment is largely dependent on the tensile force generated at the MTU, and muscle and tendon moment arms are constant (joint angle dependent), variability of muscle activation amplitudes and latency characteristics may have a strong interdependency with an external auditory cueing stimulus. A finding that rhythmical tasks that utilise the SSC at relatively high forces are affected so markedly with respect to the mechanical work done in braking phase, may be associated with the etiology of injury and improving rehabilitation strategies.

The calculation of agonist and antagonist co-contraction was based on the quotient between peak muscle activation values rather than the instantaneous ratio when the amplitude of muscle activation may be different between agonist and antagonist muscles. Therefore, future studies may aim to determine the instantaneous co-contraction ratio and whether this more accurately represents the control parameters of opposing muscle groups at a joint during rapid dynamic loading tasks. Furthermore the EMG signal may be different at different muscle lengths that may be experienced under different hopping conditions and confound the results of differences found between hopping conditions. The large variance in knee joint stiffness may be due to the calculation of stiffness over the complete braking phase rather than at peak values for either joint moment or joint excursion. Although, it is demonstrated within the literature on hopping, that during sub-maximal efforts the ankle and knee were acting in synergy and almost simultaneously during the braking phase (Auyang, et al., 2009; Blickhan, et al., 2007; Bobbert, et al., 2011; Farley, et al., 1991; Farley, et al., 1998; Farley, et al., 1999; Hobara, et al., 2008; Yen, et al., 2009a; Yen, et al., 2009b), future studies may aim to verify that this is the same pattern in the hopping conditions utilised in the current investigation as this may affect the measure of joint stiffness. This may occur as a result of differences in the rate of change for joint moment and joint excursion resulting in differences in instantaneous stiffness.

The analysis of kinematics and kinetics was only performed for motion in the sagittal plane similar to all previous studies in the area. It is possible that significant changes across conditions may occur in the frontal and transverse planes as they are shown to change in the sagittal plane. These possible changes may be clinically significant and a study of kinematics and kinetics in the frontal and transverse planes is a clear prospect for future investigation in lower limb loading activities.

3.5.6 Clinical Significance

The findings of this study are specific to the sample population tested consisting of elite level jumping athletes who participate in volleyball and basketball. As noted in the introduction, these groups of athletes are most susceptible to patellar tendinopathy that is related to the overuse of the PT (Lian, et al., 2005; Tiemessen, et al., 2008). The current findings demonstrate that under certain landing conditions the knee joint is susceptible to higher than expected forces for even a sub-maximal task such as hopping on-the-spot. Kovács et al. (1999) reported greater work for the triceps surae during a forefoot-first compared to heel-first landing from drop jumps, with no difference in EMG activity (Kovács, et al., 1999). Kovács et al. attributed these differences to a difference in stored elastic energy rather than increased muscle activity. A condition under which this may occur is when the ankle joint is mechanically compromised by not being able to generate adequate joint stiffness, moments and not being able to perform the mechanical work necessary in braking during landing as would be expected on a flat surface.

The hypothesis that reduced ankle function in athletes with knee pain is supported by a finding of flat-footed landings being more common in basketball players with patellar tendinopathy than controls (Siegmund, Huxel, & Swanik, 2008). Furthermore, reduced ankle joint function is associated with an increased in load transfer to the knee joint (Dufek & Bates, 1990; Gross & Nelson, 1988; S.-N. Zhang, Bates, & Dufek, 2000). This pattern of modulation was demonstrated in the current study with a finding of reduced mechanical work at the ankle joint and greater mechanical work done and angular excursion at the knee joint. This was most notable in the DL loading condition where there is a conservation of muscle activation amplitude of the VL. Although the current study did not test participants with knee pain, future studies need to empirically determine whether pain can influence a reduction in ankle or knee joint function that was demonstrated between different hopping surfaces.

In the sports of volleyball and basketball a number of situations may arise where ankle joint function is compromised. In both sports it may be considered that the majority of vertical jump efforts will be followed by the objective to land using a DL or step-close strategy, especially in volleyball where the field of play does not permit large horizontal movements. Descriptive data on knee flexion in landing report that the angle is greater during landings from a ‘block’ than ‘spike’ jump in male volleyball players (Salci, Kentel, Heycan, Akin, & Korkusuz, 2004). Ankle joint function may be reduced when landing flat-footed thereby reducing the capacity of the ankle joint to effectively modulate braking in landing (Kovács, et al., 1999), similar to decline hopping. This is most often dependant on the state of play in these open field sports. For example, in volleyball a player may be unable to land on the forefoot as they may be too close to the net and thereby forced to land flat-footed to avoid touching the net. This is a similar scenario to basketball where players may avoid a forefoot contact strategy in response to opposition player position and land flat-footed to avoid contact with other players. Stoffell et al. (2010) demonstrated that ankle joint taping reduced ankle joint range of motion, knee joint internal rotation and varus moments with an increase in knee joint valgus impulse, during side-stepping maneuvers, demonstrating the effect that limiting ankle joint function may have on knee joint mechanical behaviour (Stoffell, Nicholls, Winata, Dempsey, Boyle, & Lloyd, 2010). Measuring variables at each joint in the lower limb is required to reflect possible changes along the kinetic chain. This is evident in a recent study that demonstrated no difference in vertical leg stiffness between ankle taping, bracing and control conditions (Williams & Riemann, 2009) however; there may have been change in mechanical behaviour at the joint(s) that may reflect change in load. The current study demonstrates that measuring function along the kinetic chain in the lower limb is necessary when identifying factors that may predispose individuals to injury or impact the design of appropriate rehabilitation strategies.

Eccentric exercise including single leg decline squats have been demonstrated to be effective in reducing pain and symptoms related to patellar tendinopathy (Frohm, Saartok, Hallvorsen, & Renström, 2007b; Jonsson & Alfredson, 2005; Kongsgaard, Kovanen, Aagaard, Doessing, Hansen, Lauren, Kaldau, Kjaer, & Magnusson, 2009; Purdam, et al., 2004; Visnes & Bahr, 2007; Young, Cook, Purdam, Kiss, & Alfredson, 2005). The use of the decline surface for squatting was shown to increase PT load (Frohm, et al., 2007a; Kongsgaard, et al., 2006; Purdam, et al., 2003; Purdam, et al., 2004; J. G. Richards, et al., 2008; Zwerver, et al., 2007) and was the basis for performing hopping on a decline surface. The finding in the current study to preferentially load the knee extensor muscles during a repeated SSC task, especially during DL hopping on a decline board, may be a focus of future studies that aim to develop a surrogate strategy to strengthen the PT as part of a training or rehabilitation program.

3.5.7 Summary and Conclusion

The hopping procedure instituted in this study has determined that the interaction between the ankle and knee joints is complex at sub-maximal hopping efforts. The ankle joint does appear to be the primary site for modulation, with the ability to utilise the calf complex as a spring to store and utilise elastic energy (Kawakami & Fukunaga, 2006) and due to the anthropometric advantage of having its lever arm near perpendicular to the GRF in the braking phase. The knee joint is not a passive link in the kinetic chain and even at the sub-maximal efforts in this study, contributes significantly to the braking phase in hopping, especially when the ankle joint is compromised or constrained. In these cases the primary mechanical behaviours exhibited at the knee joint are of increasing the joint excursion and joint moment. Although the motor system appears to conserve knee joint stiffness, the non-contractile components may be implicated in accommodating an increase in mechanical work done at the knee joint during specific landing conditions.

4 Chapter Four: Comparison of Reference and Field Test Methods for Vertical Leg Stiffness

4.1 Chapter Aims

This chapter aims to:

- compare the values of vertical leg stiffness, vertical height displacement and peak ground reaction force between the reference and the field test method that is based on the spring-mass model.

4.2 Introduction

Vertical leg stiffness is the ratio of the change in force to the change in vertical displacement. The field test method uses the ratio of contact time (T_c) with flight time (T_f) to fit the behaviour of hopping to a simple harmonic motion. This is equivalent of a single spring with a mass at the end of the spring, i.e.; a spring-mass model. The field test method (Dalleau, et al., 2004) is based on the assumption that the GRF to time curve is sinusoidal and best describes the simple harmonic motion. Hooke's law states that the displacement of a spring by a set force is proportional to the stiffness of the spring and that a body that deforms may store and return energy based on the formula $F = kx$, where the force (F) required to deform the body is proportional to the proportionality constant (k) and the distance (x) that the object is deformed, on the condition that the body is not permanently deformed (R. J. Butler, et al., 2003). Therefore, integrating both Hooke's law and the movement model of simple harmonic motion can allow estimation of the displacement and peak force.

A number of methods have been used for calculation of vertical stiffness (R. J. Butler, et al., 2003) and in this study the reference method used describes the ratio of change in vGRF to the change in displacement of the COM during braking phase, where motion is assumed to occur only along the z axis in a linear manner (T. A. McMahon, et al., 1990). In this case, the measure of vertical stiffness is noted as being a measure of 'quasi-stiffness' as it does not model each element that contributes to stiffness including ligament, tendon, muscle, bone and cartilage as well as neural contributions in controlling a multi-joint system (M. L. Latash, et al., 1993).

Vertical stiffness is modeled as deformation of the COM during the braking phase and is termed vertical leg stiffness (Kuitunen, et al., 2002; T. A. McMahon, et al., 1990) the term used throughout this chapter. It is understood that the leg is considered the spring element and that deformation will primarily occur as a shortening of the leg. Hence, previous studies have estimated leg spring deformation and calculated ‘leg stiffness’ (Kerdok, et al., 2002; P. J. McMahon, et al., 1987; T. A. McMahon, et al., 1990; Morin, et al., 2005). Surface stiffness is shown to affect leg stiffness (Farley, et al., 1998; Ferris, et al., 1997; Ferris, et al., 1998), however in this study with a repeated measures design and no change in surface stiffness between hopping conditions, the measure of vertical leg stiffness was seen as sensitive to changes due to hopping condition only. There is also shown to be a strong correlation between vertical stiffness and leg stiffness (Morin, et al., 2005).

The purpose of this study was to examine the validity of the field test method during different hopping conditions.

4.2.1 Hypotheses

This investigation tested the hypothesis:

1. That there is no difference in the measure of vertical leg stiffness, vertical height displacement or peak vGRF between the reference and field test methods.

4.3 Methods

Data were collected and analysed as described in chapter 3 (section 3.3) for the 9 participants included in the study of lower limb motor control during hopping. Briefly, the factors of hopping pace (self-selected and fixed (1.8Hz)), task (DL and AL hopping) and surface (flat or decline) were changed between trials in a series of eight hopping conditions. The values obtained for vertical leg stiffness, force and vertical height displacement during braking phase, were compared between reference and field test methods of calculation.

The reference method is the gold standard for calculation of force, vertical height displacement and vertical leg stiffness as it is based on Newton's Laws of motion (Watkins, 2007).

4.3.1 Reference Method

From first principles, the displacement of the COM from IC to peak vGRF is estimated by the action of gravity acting on the mass during the second half of flight phase. The time in the flight phase is known by identifying the toe-off of one hop and IC of the successive hop. The vertical height displacement to time curve is parabolic and at maximum height in flight phase, velocity is 0m/s. This allowed a calculation of time in the descending phase of flight that is half of the time in flight between successive hops.

$$z_f = \frac{1}{2} \cdot g \cdot \left(\frac{T_f}{2}\right)^2 \qquad \text{.......equation 1}$$

Where,

z_f is the vertical displacement of a mass from peak height to IC (second half of flight phase), g is acceleration due to gravity (9.81m.s^{-2}) and T_f is time in flight (from peak height to IC).

The velocity at the point of IC is estimated by the acceleration due to gravity

$$v_i = \sqrt{2.g.z_f} \qquad \text{.......equation 2}$$

Where,

v_i is the velocity at IC, g is acceleration due to gravity (9.81m.s^{-2}) and z_f is the vertical displacement of the mass from the peak height to IC.

Then, the displacement of the COM from IC to peak depression (braking phase) is determined by

$$z_c = \frac{1}{2}\left(v_i + v_f\right) \cdot \left(\frac{T_c}{2}\right) \qquad \text{.......equation 3}$$

Where,

z_c is the vertical displacement of the mass from IC to peak vGRF (braking phase), v_f is the final velocity at peak vGRF (0 m.s^{-1}) and T_c is time in contact.

Vertical leg stiffness ($N.m^{-1}$) k_R was then derived.

$$k_R = \frac{\Delta vGRF}{z_c} \qquad \text{……..equation 4}$$

4.3.2 Field Test Method

The field test method describes the calculation of vertical leg stiffness using the measured values of flight and contact time only (Dalleau, Belli, Bourdin, & Lacour, 1998) and assumes the vGRF to behave in a sinusoidal fashion (Dalleau, et al., 1998). The equation used to calculate vertical leg stiffness using the field test method (k_D) is the ratio of peak force to total displacement at half contact phase (Dalleau, et al., 2004). The expressions used to calculate peak force (F_{max}) and total height displacements (z) are shown below.

$$F_{max} = Mg \times \frac{\pi}{2} \times \left(\frac{T_f}{T_c} + 1 \right) \qquad \text{……equation 5}$$

$$z\left(\frac{T_c}{2}\right) = \frac{-F_{max}}{M} \times \left(\frac{T_c^{\ 2}}{\pi^2}\right) + g \cdot \frac{T_c^{\ 2}}{8} \qquad \text{……equation 6}$$

Where,

M is the mass of the participant, g is acceleration due to gravity ($9.81\ m.s^{-2}$), T_f is flight time and T_c is contact time

Vertical leg stiffness ($N.m^{-1}$) (k_D) was then derived.

$$k_D = \frac{F_{max}}{z} \qquad \text{…….equation 7}$$

4.3.3 Data Analyses

Correlation and linear regression analyses were performed to compare the vertical leg stiffness, peak force and vertical height displacement measures in braking phase, calculated by the reference method (R) and the field test method (D). Each pair of values calculated from the two methods was entered into a spreadsheet (W.G. Hopkins, 2000) for each of the 8 hopping conditions. Linear correlations were identified and the mean and 95% CI of the regression coefficient (slope) and Y intercept for vertical leg stiffness, peak force and vertical height displacement were determined. Typical error values (SDdiff/$\sqrt{2}$, where SDdiff is the standard deviation of the difference scores) and 95% CI were determined for the paired comparisons of *k*, F and z (W.G. Hopkins, 2000).

4.4 Results

The main finding was that for vertical leg stiffness, the Pearson's correlation coefficients were very high (>0.9) for DL hopping on the flat surface at a self-selected pace (2.0Hz) and forced pace (1.8Hz) and DL hopping on the decline surface at a forced pace (1.8Hz) with typical errors ranging from 1.8 to 2.5kN.m^{-1} of the reference method (Table 4.4.1). Vertical leg stiffness values were underestimated using the field test method across all hopping conditions (Table 4.4.1).

For peak vGRF Pearson's correlation co-efficient values were hgih ranging from 0.79 to 0.99 across all hopping condition (Table 4.4.2). Peak vGRF was underestimated using the field test method across all hopping conditions. The correlation co-efficient values for vertical height displacement were also very hgih ranging from 0.67 to 0.95 (Table 4.4.3). Vertical height displacement was overestimated using the field test method.

Derived variables for a single participant during AL hopping on a decline surface at a self-selected pace was unable to be calculated as he had no flight phase during this task and therefore did not hop.

Table 4.4.1 Comparison of vertical leg stiffness ($kN.m^{-1}$) (mean (SD)) with values for the Pearson's correlation co-efficient (R) and linear regression analysis including the slope [95%CI], intercept [95%CI] and typical error [95%CI].

Hopping Condition	Reference	Field Test	R	Slope [95% CI]	Intercept [95% CI]	Typical Error [95%CI]
DL Flat S	44.2 (9.6)	22.3 (5.6)	0.97	1.6 [1.3 to 1.9]	7.4 [0.6 to 14.3]	2.5 [1.8 to 4.5]
DL Dec S	40.8 (5.8)	20.2 (3.5)	0.68	1.1 [0.3 to 2.0]	18.0 [0.4 to 35.7]	4.5 [3.2 to 8.2]
AL Flat S	115.0 (69.8)	21.6 (3.1)	0.32	7.2 [-8.0 to 22.4]	-39.9 [-370.6 to 290.9]	70.7 [49.9 to 127.0]
†AL Dec S	90.1 (38.1)	21.9 (2.8)	0.60	8.2 [-1.7 to 18.0]	-89.2 [-306.6 to 128.1]	37.7 [25.4 to 78.8]
DL Flat F	42.1 (10.7)	23.0 (7.8)	0.98	1.3 [1.1 to 1.5]	11.3 [6.7 to 16.0]	2.3 [1.6 to 4.1]
DL Dec F	32.1 (4.9)	18.6 (3.9)	0.94	1.2 [0.9 to 1.5]	10.4 [4.5 to 16.2]	1.8 [1.3 to 3.3]
AL Flat F	35.5 (13.3)	14.3 (2.2)	0.14	0.9 [-3.5 to 5.2]	23.1 [-40.1 to 86.3]	14.1 [10.0 to 25.4]
AL Dec F	36.3 (23.5)	13.7 (2.1)	0.41	4.6 [-2.7 to 11.8]	-26.7 [-127.0 to 73.6]	22.9 [16.1 to 41.1]

† - n = 7, DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace, F – Forced Pace, Dec – Decline

Table 4.4.2 Comparison of peak force (kN) (mean (SD)) in braking phase with values of the Pearson's correlation co-efficient (R) and linear regression analysis including the slope [95%CI], intercept [95%CI] and typical error [95%CI].

Hopping Condition	Reference	Field Test	R	Slope [95% CI]	Intercept [95% CI]	Typical Error [95%CI]
DL Flat S	3.5 (0.8)	2.7 (0.7)	0.98	1.2 [1.1 to 1.4]	0.1 [-0.3 to 0.60]	0.2 [0.1 to 0.3]
DL Dec S	2.7 (0.5)	2.3 (0.4)	0.99	1.1 [1.0to 1.3]	0.1 [-0.2 to 0.4]	0.1 [0.16 to 0.2]
AL Flat S	2.1 (0.3)	1.6 (0.2)	0.79	1.3 [0.6 to 2.0]	-0.03 [-1. 2 to 1.2]	0.2 [0.1 to 0.3]
†AL Dec S	1.9 (0.3)	1.6 (0.2)	0.97	1.1 [0.8 to 1.3]	0.2 [-0.2 to 0.6]	0.1 [0.1 to 0.2]
DL Flat F	3.7 (0.7)	3.1 (0.7)	0.98	1.0 [0.80 to 1.1]	0.7 [0.2 to 1.2]	0.2 [0.1 to 0.3]
DL Dec F	2.9 (0.5)	2.7 (0.5)	0.99	1.0 [0.9 to 1.1]	0.2 [-0.2 to 0.5]	0.1 [0.1 to 0.2]
AL Flat F	2.2 (0.4)	2.0 (0.3)	0.94	1.0 [0.7 to 1.3]	0.3 [-0.3 to 0.8]	0.1 [0.1 to 0.2]
AL Dec F	1.9 (0.2)	1.9 (0.3)	0.91	0.8 [0.5 to 1.1]	0.4 [-0.1 to 0.9]	0.1 [0.1 to 0.2]

† - n = 7, DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace, F – Forced Pace, Dec – Decline

Table 4.4.3 Comparison of vertical height displacement (cm) (mean (SD)) in braking phase with values of the Pearson's correlation co-efficient (R) and linear regression analysis including the slope [95%CI], intercept [95%CI] and typical error [95%CI].

Hopping Condition	Reference	Field Test	R	Slope [95% CI]	Intercept [95% CI]	Typical Error [95%CI]
DL Flat S	8.0 (1.8)	12.7 (2.7)	0.95	0.6 [0.5 to 0.8]	0.1 [-1.9 to 2.1]	0.6 [0.4 to 1.1]
DL Dec S	6.6 (1.4)	11.5 (1.7)	0.83	0.7 [0.4 to 1.0]	-1.3 [-5.0 to 2.5]	0.8 [0.6 to 1.5]
AL Flat S	2.5 (1.4)	7.7 (1.5)	0.67	0.6 [0.1 to 1.1]	-2.2 [-6.1 to 1.6]	1.1 [0.8 to 2.0]
†AL Dec S	2.6 (1.4)	7.5 (1.5)	0.88	0.8 [0.4 to 1.2]	-3.5 [-6.5 to -0.6]	0.7 [0.5 to 1.5]
DL Flat F	9.0 (0.8)	13.7 (1.6)	0.91	0.5 [0.3 to 0.6]	2.5 [0.4 to 4.5]	0.4 [0.3 to 0.6]
DL Dec F	9.2 (0.5)	14.3 (0.8)	0.76	0.8 [0.3 to 1.3]	-2.4 [-9.4 to 4.6]	0.8 [0.3 to 1.3]
AL Flat F	7.5 (1.1)	13.8 (1.2)	0.86	1.3 [0.7 to 1.9]	-11.3 [-19.1 to -3.5]	1.0 [0.7 to 1.9]
AL Dec F	6.9 (1.6)	13.7 (1.7)	0.86	1.1 [0.6 to 1.6]	-9.1 [-15.6 to -2.6]	1.2 [0.9 to 2.2]

† - n = 7, DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace, F – Forced Pace, Dec – Decline

4.5 Discussion

The main finding was that the field test method provides a good estimate for the measure of vertical leg stiffness during DL hopping on a flat surface at a self-selected pace (2.0Hz) and for DL hopping on flat and decline surfaces at a forced pace (1.8Hz). The field test method models the behaviour of the human body during the hopping skill as a spring-mass with the GRF modeled as a sine wave and the force-time integral assumed to be the impulse during contact (Dalleau, et al., 2004). Therefore, the ratio of peak GRF and vertical height displacement in contact provides an estimate of vertical leg stiffness.

Estimates of peak force using the field test method were good across all eight hopping conditions ($r > 0.79$) with the typical error ranging from 0.1kN [95%CI] [0.1 to 0.2] to 0.2kN [0.1 to 0.3]. This demonstrates that the major source of error using the field test method for calculation of vertical leg stiffness was the estimate of vertical height displacement rather than of force. Although correlations for vertical height displacement were good (>0.7) the typical error ranged from 0.4cm [95%CI] [0.3 to 0.6] to 1.2cm [0.9 to 2.2] of the estimates calculated using the reference method. The overestimate of vertical height displacement is supported by a previous study on running (Bullimore & Burn, 2007).

The current finding supports a previous investigation that examined the validity of using the field test method against sub-maximal DL hopping (2.0 and 2.5Hz) demonstrating low typical errors of 6.5% and 7.5% for estimates of vertical leg stiffness in each of these tasks respectively (Lloyd, et al., 2009). This may be interpreted as a finding that these hopping conditions are well modeled as a spring-mass and that the GRF is described by the sine wave (Blickhan, 1989). However, for estimates of vertical leg stiffness during other hopping conditions, especially AL hopping, at 2.4Hz and 1.8Hz, the field test method does not accurately estimate vertical leg stiffness with Pearson's correlation coefficient values ranging from 0.1 to 0.7.

The findings in the current hopping tasks are supported by findings during running at different speeds on a treadmill (3.33 to 6.67$m.s^{-1}$) and overground (4 to 7$m.s^{-1}$ and maximal velocity) (Morin, et al., 2005). There are very strong correlations between the reference method used (Cavagna, 1975) and a spring-mass model, based on the field test method for both vertical stiffness (k_{vert}) (ratio between F_{max} and vertical displacement of the COM (Δy_c)) and leg stiffness (k_{leg}) (ratio between F_{max} and the change in length of leg spring (ΔL) (see below) (Morin, et al., 2005).

$$\Delta L = L - \sqrt{L^2 - \left(\frac{vt_c}{2}\right)^2} + \Delta y_c$$

Where L is the leg length at standing (L), running velocity is in m.s^{-1} and time in contact (t_c) is in s.

Morin et al. (2005) demonstrated that k_{vert} and k_{leg} were underestimated during overground running by a mean of 2.3% (SD) (1.63) and 2.54% (1.16) respectively. Results were similar for these two variables during treadmill running with a mean of 0.12% (SD) (0.53) and 6.05% (3.02). Correlations between k_{vert} and k_{leg} at either running condition were strong ranging from 0.89 to 0.98, supporting the current method of calculating vertical leg stiffness as a ratio of the change in vGRF to vertical displacement of the COM. Modeling running at different velocities and DL hopping (at near preferred frequencies on a flat or decline surface) are well represented by the GRF behaving in a sinusoidal fashion (Bullimore, et al., 2007). However, it must be noted that the values are underestimated. If behaviour is not in a sinusoidal fashion then there may be difficulty in interpreting the results for an individual or sample population without a reference method for comparison.

Measures of vertical leg stiffness at maximal height hopping were similar to the current findings with the typical error of 41.9% (Lloyd, et al., 2009). It has previously been demonstrated that during DL hopping at frequencies lower (1.2, 1.5 and 1.8Hz) than the preferred frequency of 2.2Hz, the system did not behave like a spring-mass system and that the preferred frequency is based on the time needed to maximise force generation in contact (Hobara, et al., 2010a). A number of mechanisms have been proposed to contribute to the behaviour of a human hopping to a spring-mass, primarily based on the physiological parameters of reducing metabolic cost and optimisation of rate of force development at a preferred hopping frequency (Blickhan, 1989; Lloyd, et al., 2009).

In determining vertical leg stiffness during human hopping it is necessary to validate the use of the spring-mass model based on the task specific conditions of hopping including the loading strategy (single- or double-leg), surface (flat or decline) and hopping frequency (self-selected or forced). If these conditions violate the assumptions of a spring-mass model, in that the GRF must be modeled as a sine wave, then it is necessary to use a gold standard measure that is recording vGRF and calculating vertical height displacement during contact period using first principles. A limitation of this analysis is the assumption that the association between the two test methods is linear. However, due to the small sample size it was not possible to determine the validity of the linear fit model.

4.5.1 Summary and Conclusion

The findings in this study justify the use of a first principles approach of measuring vGRF and calculating the vertical height displacement in braking phase based on first principles of motion and force. The spring-mass model did not represent the behaviour of participants when hopping using an AL hopping strategy. In conclusion there are difficulties related to the hopping conditions under which the field test method may be used to calculate vertical leg stiffness. The field test method is strongly correlated with the reference method when DL hopping on a flat surface at either 2.0Hz or 1.8Hz and DL hopping on a decline surface at 1.8Hz.

5 Chapter Five: *In vivo* Patellar Tendon Mechanical Properties

A Pilot Study

5.1 Chapter Aims

This chapter aims to:

- evaluate the validity of measuring patellar tendon elongation *in vivo*.
- evaluate the reliability of measuring patellar tendon mechanical properties *in vivo*.

5.2 Introduction

The focus of this study was to establish validity of measuring *in vivo* PT length and reliability of measuring PT mechanical properties (strain, stiffness, stress and elastic modulus) *in vivo*. Calculation of PT mechanical properties required the measure of PT elongation, force and cross sectional area (CSA).

In vivo diagnostic (B-mode) ultrasound imaging (Gibbon, 1996; Whittingham, 2007) of the PT is a non-invasive and low-cost means to measure PT elongation during knee extensor contraction and has been used extensively in previous studies (P. Hansen, Bojsen-Møller, Aagaard, Kjaer, & Magnusson, 2006; Kongsgaard, Reitelseder, Pedersen, Holm, Aagaard, Kjaer, & Magnusson, 2007; Kubo, et al., 2005b; Onambélé, et al., 2007a; S. P. Pearson, Burgess, & Onambélé, 2007; N. D. Reeves, Maganaris, & Narici, 2003a; Westh, et al., 2008). Inherent difficulties in utilising ultrasound are low sampling frequency and stability of the contact between the transducer and skin (Hoskins, 2003). Other studies have not visualised both proximal and distal PT attachments and aimed to quantify the error associated with only measuring either the proximal or distal PT segment lengths (Burgess, Pearson, Breen, & Onambélé, 2009b; Burgess, Pearson, & Onambélé, 2009c; Kongsgaard, et al., 2006; Kubo, Yata, Kanehisa, & Fukunaga, 2006; Maganaris, Narici, & Reeves, 2004; S. J. Pearson & Onambélé, 2006; N. D. Reeves, et al., 2003a). The available ultrasound device in the current study did not allow visualisation of the complete PT length on a single field of view during ramp maximal voluntary isometric contraction (MVIC) of the knee extensors in taller individuals. Hence, measuring elongation of the complete PT length remains problematic and validating a method to accurately measure complete PT elongation was the focus of this study. This pilot study recruited participants for whom the complete PT length could be visualized and aimed to establish a reliable and accurate method to measure of PT mechanical properties. The aim was to validate a method for measuring PT mechanical properties in elite level jumping and non-jumping athletes for whom the complete PT length may not be able to be visualised in a single field of view.

5.2.1 Patellar Tendon Elongation *In Vivo* using Ultrasound

Previously, to establish a frame of reference for the proximal segment (Burgess, et al., 2009c; Kongsgaard, et al., 2006; Kubo, et al., 2006; Maganaris, et al., 2004; S. J. Pearson, et al., 2006; N. D. Reeves, et al., 2003a) or both proximal and distal segments (Burgess, et al., 2009b) and to act as a reference for movement of the transducer relative to the skin, a marker adhered to the skin over the mid PT length has been utilised. The validity of these techniques relies on the assumption that the frame of reference of the shadow of the skin marker on the ultrasound image is the same as the frame of reference of the bony landmark. If this assumption is true then the proportion of elongation of the segment between the bony insertion and shadow should reflect the proportional elongation of the total PT length. In other words the shadow marker remains at a proportional distance between the two bony landmarks during the changes in load during the testing. Clearly an error in the assessment of PT elongation may occur if the shadow of the marker moves independently to that of the bony tendon attachment. Hansen et al. (2006) reported findings of their pilot data which showed that accounting for proximal PT deformation only, was inaccurate in estimating PT length deformation due to movement of the tibia distally (P. Hansen, et al., 2006) and that it is necessary to account for the deformation of the complete length of the PT (Bojsen-Møller, Brogaard, Have, Stryger, Kjaer, Aagaard, & Magnusson, 2007). Pearson et al. (2007) also suggested that not accounting for motion at the tibia does not account for creep along the full PT length in their comparison of PT deformation at two speeds of knee extensor contraction (S. P. Pearson, et al., 2007). No studies have investigated the validity of comparing change of the total PT length against the sum of proximal and distal PT segment lengths, proximal length only or distal segment length only.

5.2.2 Patellar Tendon Moment Arm Length

Patellar tendon moment arm length has previously been estimated using a number of methods. Direct methods rely on identification of anatomical landmarks that define a knee axis of rotation and the PT that attaches proximally at the inferior pole of the patella and distally at the tibial tuberosity. Thus, a geometric measure may be taken (Tsaopoulos, Baltzopoulos, & Maganaris, 2006a) as the distance perpendicular distance for the line of action of the PT to the knee joint axis of rotation. This principle has been used from imaging techniques involving x-ray (Smidt, 1973), magnetic resonance imaging (MRI) (Maganaris, Baltzopoulos, & Tsaopoulos, 2005; Maganaris, et al., 2004; O'Brien, Reeves, Baltzopoulos, Jones, & Maganaris, 2010; N. D. Reeves, et al., 2003a; N. R. Reeves, Maganaris, Maffulli, & Rittweger, 2009; Sheehan, 2007; Tsaopoulos, Maganaris, & Baltzopoulos, 2006b; Wretenberg, Nemeth, Lamontagne, & Lundin, 1996) and computed tomography video fluoroscopy (Baltzopoulos, 1995; Kellis & Baltzopoulos, 1999b). Investigation of the relationship between anthropometric measures and PT moment arm length were shown to have poor concordance (low correlation co-efficient) (Tsaopoulos, et al., 2006b) and necessitated the development of a method of reliably measuring PT moment arm length directly.

An indirect method was the tendon excursion method of estimating PT moment arm length and involved applying a known force in the line of the knee extensor MTU and simultaneously measuring the joint angular rotation. As the applied force, joint moment and angular rotation are known, the PT moment arm length may be calculated (An, Takahashi, Harrigna, & Chao, 1984; Spoor & van Leeuwen, 1992). This method has a number of constraints in that the limb must be in motion and hence, this method has traditionally been used as an *ex vivo* method on cadavers (Buford, Ivey, Malone, Patterson, Peare, Nguyen, & Stewart, 1997).

Numerous studies to date have reported varying ranges of PT moment arm length and are dependent on the knee angle of flexion. Of specific interest to this study was the PT moment arm length estimate at 90º knee flexion that has been reported to vary from 20 to 50mm (Baltzopoulos, 1995; Herzog & Read, 1993; Krevolin, Pandy, & Pearce, 2004; Tsaopoulos, et al., 2006a). A number of reasons may explain the large variance in results. Primarily the method of estimate varies between each of the studies, as the direct geometric measure is dependent on the accuracy of the imaging (MRI) to represent the knee anatomy. Other ‘indirect techniques’ introduce systematic errors in the measure of force and knee joint lever arm length estimates. This latter point also introduces a key disparity that exists in the literature that is related to the axis of rotation of the knee. Commonly used anatomical reference points are the tibio-femoral contact point (Maganaris, et al., 2005; Nisell, Nemeth, & Ohlsén, 1986) and the cruciate ligament cross point, both of which have been used as an estimate of knee axis of rotation and consequently used to measure the perpendicular distance to the line of pull of the PT (Gill & O'Connor, 1996). The choice of these points is largely dependent on the method used to derive the PT moment arm length rather than an assumption that either is the instantaneous axis of rotation (Tsaopoulos, et al., 2006a).

Considering the inherent difficulties and errors in measuring PT moment arm length, a quick and cost-effective device based on the direct anthropometric measurement of the perpendicular distance from the line of pull of the PT to the knee axis of rotation that is the transepicondylar axis has been described (Churchill, Incavo, Johnson, & Beynnon, 1998; Hollister, Jatana, Singh, Sullivan, & Lupichuk, 1993; Matsuda, Miura, Nagamine, Urabe, Mawatari, & Iwamoto, 2003; Stiehl & Abbott, 1995; Yoshino, Takai, Ohtsuki, & Hirasawa, 2001). Many authors have described the fact that the transepicondylar axis (a direct line between the medial and lateral femoral epicondyles) is the closest approximation to the knee axis of rotation (Asano, Akagi, & Nakamura, 2005; Berger, Rubash, Seel, Thompson, & Crossett, 1993; Churchill, et al., 1998; Hollister, et al., 1993; Smidt, 1973; P. N. Smith, et al., 2003; Stiehl, et al., 1995). The lateral and medial femoral epicondyles that define the transepicondylar axis are easily palpable and routinely used as a stable reference point during surgery and utilised in aligning the femur correctly during total knee arthroplasty (Akagi, Yamashita, Nakagawa, Asano, & Nakamura, 2001; Berger, et al., 1993; Luo, Koshino, Takeuchi, & Saito, 2001; Stiehl, et al., 1995; Yoshino, et al., 2001; T. Yoshioka, Siu, & Cooke, 1987).

To minimise the potential measurement error identifying the transepicondylar axis of rotation, line of pull of the PT and the inferior pole of the patella a custom-built caliper was developed. The secondary aim of this study was to evaluate the reliability of measuring PT mechanical properties where the measure of PT moment arm length using the custom-built caliper was used to allow calculation of the PT force.

5.2.3 Patellar Tendinopathy

The proximal and posterior portion of the PT is the common site for patellar tendinopathy (Cook, et al., 2001a; Gemignani, et al., 2008; Hoksrud, et al., 2008; Witvrouw, et al., 2001). The proximal portion of the PT has been shown to have an increased strain posteriorly (Lavagnino, et al., 2008) and greater stress than other portions of the PT (Dillon, et al., 2008). In contrast the posterior portion of the PT has also been shown to have a decreased strain compared to the anterior portion in knee flexion (Almekinders, et al., 2002). The proximal PT is also shown to have a different CSA than other points along the length of the PT. These differences may be associated with the type of sport that individuals participate in (Couppé, Kongsgaard, Aagaard, Hansen, Bojsen-Møller, Kjaer, & Magnusson, 2008; Kongsgaard, et al., 2007; Westh, et al., 2008). It is also reported that PT strain is common at the bony attachments and higher than the mid-region (D. L. Butler, Sheh, Stouffer, Samaranayake, & Levy, 1990). Therefore, a number of physiologic parameters that are specific to the proximal and posterior portion of the PT are of interest in establishing a method to evaluate PT CSA and stress. This study aimed to determine the reliability of determining elongation specific to the posterior PT and PT CSA and stress specific to the proximal PT.

The primary aim of this study was to establish an accurate method to measure total PT length by comparing the gold standard of measuring the total length (TL) to measuring the PT length as a sum of proximal (d1) and distal (d2) segment lengths. The secondary aim was to establish the reliability of measuring PT mechanical properties. This included establishing reliability of measuring PT force as derived from the measure PT moment arm length with a custom-built caliper.

5.2.4 Hypotheses

This investigation tested the following hypotheses:

1. That there is no difference in PT elongation when measured as the total PT length or as the sum of proximal (d1) and distal (d2) segment length.
2. That there is no difference in strain, stiffness, stress, elastic modulus or PT moment arm length between testing sessions.

5.3 Methods

5.3.1 Participants

Four participants (3 male and 1 female) of mean 29.8 (SD) (8.5) years of age, 170.9 (10.2) cm height and 65.3 (8.3) kg weight, were recruited from the staff and athlete population at the Australian Institute of Sport. Only one participant was an elite level athlete (male gymnast) while the other three recruits were healthy and recreationally active. The participants were skeletally mature and chosen based on the ability to visualise the complete PT length (from its bony origin to insertion) on the ultrasound viewer during a ramped MVIC of the knee extensors with the knee flexed to 90º.

Exclusion criteria were any condition that would prevent the participant from performing the testing procedure, history of knee pain, injury or surgery. This investigation was approved by the La Trobe University Human Research Ethics (Appendix A) Committee and the Australian Institute of Sport Research Ethics Committee (Appendix B). Participants were informed and provided written consent (Appendix C).

The procedures to measure PT moment arm length, PT CSA, knee moment and PT elongation were performed on both left and right legs of all participants during one session. All procedures were repeated in a second session 30 minutes later on the same day. Only a short rest period of 30 minutes was allowed as PT stiffness is observed to change based on the time of day when measures were ~14 hours apart between testing sessions (S. J. Pearson, et al., 2005). Participants were allowed to walk around and sit but not participate in any other activity.

5.3.2 Procedure

5.3.2.i Patellar Tendon Moment Arm

The participant was positioned in crook-lying with the knee flexed between 90º to 100º. The custom-made caliper was aligned to the transepicondylar axis of the knee (Churchill, et al., 1998) and the front arm at the level of the inferior pole of the patella (Figure 5.3.1). The distance between the front arm of the lever and the inferior pole of the patella was measured. This measure was taken on left and right knees for all participants.

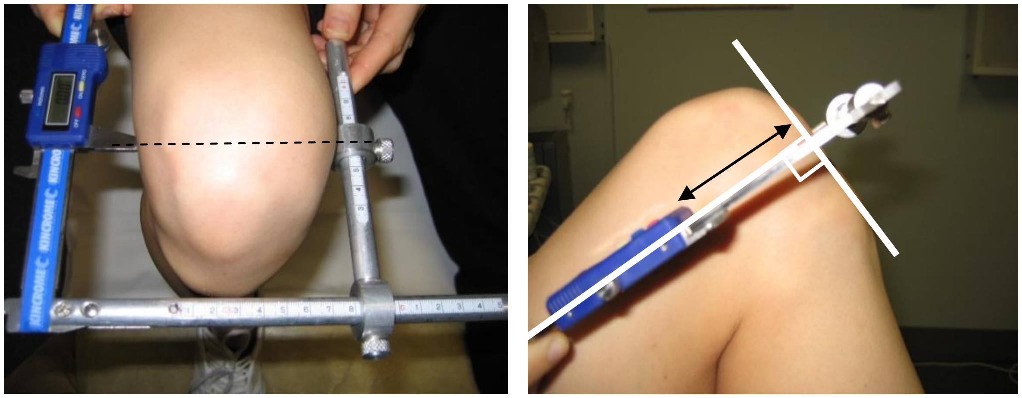


Figure 5.3.1. The custom-made caliper is aligned to the transepicondylar axis of rotation at the knee (dashed black line in left hand image). The measuring arm is perpendicular to the line of pull of the patellar tendon (solid white lines in right side image). The patellar tendon moment arm is shown by the double-ended black arrow (right hand image).

The force (N) transmitted through the PT (PT Force) during MVIC of the knee extensor muscle group was determined by dividing the knee joint moment by the PT moment arm length (Pandy, 1999; N. D. Reeves, et al., 2003a; Tsaopoulos, et al., 2006a; Wilson & Sheehan, 2009).

<u>5.3.2.ii Patellar Tendon Cross-Sectional Area</u>

Patellar tendon CSA was assessed at the proximal bony attachment of the PT (Burgess, Graham-Smith, & Pearson, 2009a; Burgess, et al., 2009b; O'Brien, et al., 2010; Onambélé, et al., 2007a; N. D. Reeves, et al., 2003a). This was achieved by recording a real-time ultrasound (Diagnostic ultrasound system, Nemio, Model SSA-550A, Toshiba, Japan) axial image of the PT, using a 12MHz, 100mm linear array transducer (PLM-805AT, Toshiba, Japan) recorded on digital video (Sony DCR-TRV900E, Japan) (Figure 5.3.2). The transducer was placed over the PT such that the edge of the transducer abutted the bony attachment of the PT to the inferior pole of the patella. This procedure was followed on left and right sides. The use of ultrasound is shown to be well correlated to the use of MRI to determine the CSA of the PT at the proximal, mid and distal portion of the PT (ICC > 0.98) (N. R. Reeves, et al., 2009).

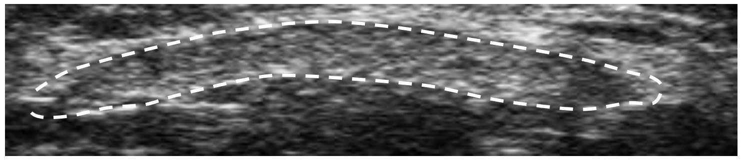


Figure 5.3.2 Axial ultrasound image of the patellar tendon (PT) at the proximal site. The top of the image is the anterior aspect of the PT. The white line is the tracing around the border of the patellar tendon and the area within this trace is the measure of the cross-sectional-area.

5.3.2.iii Patellar Tendon Elongation and Force

Patellar tendon elongation and force were determined with participants seated on a fixed dynamometer unit (KinCom, Chattanooga Corporation, Chattanooga, Tenn., USA) with the knee flexed to 90º (P. Hansen, et al., 2006; Maganaris, et al., 2004; Onambélé, et al., 2007a; S. J. Pearson, et al., 2005; S. P. Pearson, et al., 2007) and aligned to the lever arm axis of rotation (Kongsgaard, et al., 2007). Stabilisation was provided by straps about the waist, thigh and lower extremity with the arms held across the chest similar to previous studies (Kongsgaard, et al., 2007; S. P. Pearson, et al., 2007). An echo-absorptive marker that was 1.6mm in diameter (Leuko sports tape) was similar to that used in previous studies (Burgess, et al., 2009c; Kongsgaard, et al., 2006; Kubo, et al., 2006; Maganaris, et al., 2004; S. J. Pearson, et al., 2005; N. D. Reeves, et al., 2003a) and adhered to the skin at the mid-PT site between the inferior pole of the patella and tibial tuberosity (S. J. Pearson, et al., 2005; N. D. Reeves, et al., 2003a) (Figure 5.3.3).

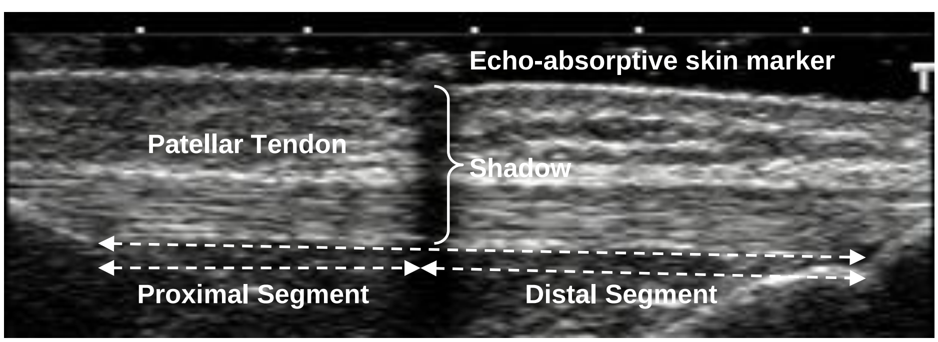


Figure 5.3.3 The echo-absorptive marker (1.6mm in diameter) was adhered to the skin at the mid-patellar tendon site between the inferior pole of the patella and tibial tuberosity. The shadow from the skin marker is shown across the mid-patellar tendon region. The complete posterior portion of the patellar tendon length is visualised (dashed line with double-ended arrow).

An ultrasound transducer (12MHz, 100mm linear array transducer, PLM-805AT, Toshiba, Japan) with a 50mm field of view, was placed in a custom-made holder that was attached by sports tape (Leuko sports tape) to the knee in the saggital plane.

This pilot study aimed to validate the method of measuring the proximal (d1) and distal (d2) segments separately and summing their values (D) and comparing this to the total length measure (TL) at common PT forces during a MVIC of the knee extensor muscle group. Hence, three recordings (d1, d2 and TL) each during a separate MVIC effort were made (Figure 5.3.4 a, b & and c). Recordings of the trials for d1, d2 and TL were made in a random order. Common to recordings of d1 and d2 was the echo-absorptive marker.

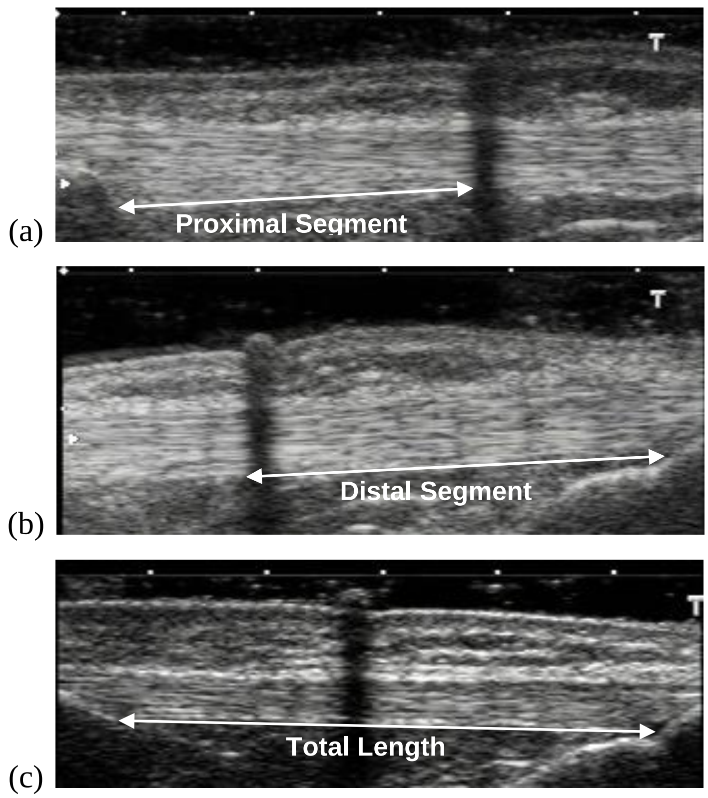


Figure 5.3.4 The proximal patellar tendon (PT) segment from the inferior pole of the patella to the proximal edge of shadow of the echoabsorptive marker (double ended white arrow) (a). The distal PT segment from the tibial tuberosity to the proximal edge of the shadow of the echoabsorptive marker (double ended white arrow) (b). The total PT length from the inferior pole of the patella to the tibial tuberosity (c).

A warm-up and familiarisation period consisting of a minimum of 4 contractions, consistent with previous studies (Einat & Yoram, 2009; Magnusson, et al., 2008; S. J. Pearson, et al., 2006; N. D. Reeves, et al., 2003a), of increasing effort of up to 100% effort were performed.

The gradual build up in effort, with rests between contractions, allowed an improvement in performance by creating a graduated increase in force and a warm-up effect by allowing an increase in effort between contractions.

The participant then performed a ramped MVIC consistent with previous studies (Couppé, et al., 2008; Kongsgaard, et al., 2007; Westh, et al., 2008). The duration of contraction was maintained as the same time period for each participant. Three of the 4 participants performed a contraction of approximately 10s duration and 1 participant performed a contraction of approximately 5s duration. The within-subject contraction duration was maintained for the same time for each participant as it has been shown that the length of duration can alter the measure of PT mechanical properties (S. P. Pearson, et al., 2007). The one participant that chose a shorter duration of hold was based on their report of comfort level in the warm-up and familiarisation period. This was deemed appropriate for this study as it was a within-subject design between sessions and a common duration of contraction was maintained across testing sessions. Each participant was instructed to develop force slowly and gradually with the instructor counting up from 1 to 10, or 1 to 5 (one participant), with reference to a digital clock, to standardise the length of time of contraction across trials.

Visual and verbal feedback was provided to ensure graduated force production. Between 1 and 4 MVICs were performed and recorded for each of d1, d2 and TL, with a minimum rest period of two minutes between efforts (Kongsgaard, et al., 2007). A random order was used for testing the left and right legs and with d1, d2 and TL for each leg.

Force data (volts) were recorded at 1000Hz on a laptop computer (Dell Precision M20, Malaysia) using data logger software (NI Data Lite Logger Version 8.0, Texas, USA) and exported into a spreadsheet (Microsoft Excel 2003) for later analysis. Ultrasound imaging was recorded on digital video (Sony DCR-TRV900E, Japan) at 25 frames per second (fps). Force and digital video data were time synchronised utilising a custom manual trigger that generated a graphic display on the ultrasound screen and created a square pulse on the analogue force trace simultaneously.

5.3.3 Data Processing

5.3.3.i Patellar Tendon Force

Prior to data collection on participants, calibration of the KinCom dynamometer analogue force output to a known mass was performed allowing conversion of the recorded trace (volts) to a known force (Newton) via the use of a linear regression equation (Appendix L).

Knee moment was calculated by the following equation:

Knee extensor moment (N.m) = Force (N) x Lever arm length (m)

PT force was calculated by the following equation:

PT Force (N) = Knee extensor moment (N.m) / PT moment arm length (m)

For each participant there were between 1 and 4 trials of MVICs recorded for each leg for d1, d2 and TL ultrasound recordings. The following criteria were used to determine the single trial from those collected for recording of d1, d2 and TL.

1. The trial with the force-time curve that had the least 'noise' and smoothest, linear increase in gradient for each of d1, d2 and TL.
2. The MVIC value for each trial was chosen at the peak of the end of the smooth part of the initial increase in voluntary force production and not in the 'noisy' region of the curve that occurs at peak efforts (Figure 5.3.5). This was performed with a visual estimation of the force-time curve and the identification of the highest point on the slope that had a positive gradient. This method may reduce the validity in being able to detect the MVIC compared to a computational method. However, this was deemed appropriate as the lowest of the three MVICs (one for each recording of the proximal, distal and total PT length) was chosen to allow force matching, rather than the best effort.

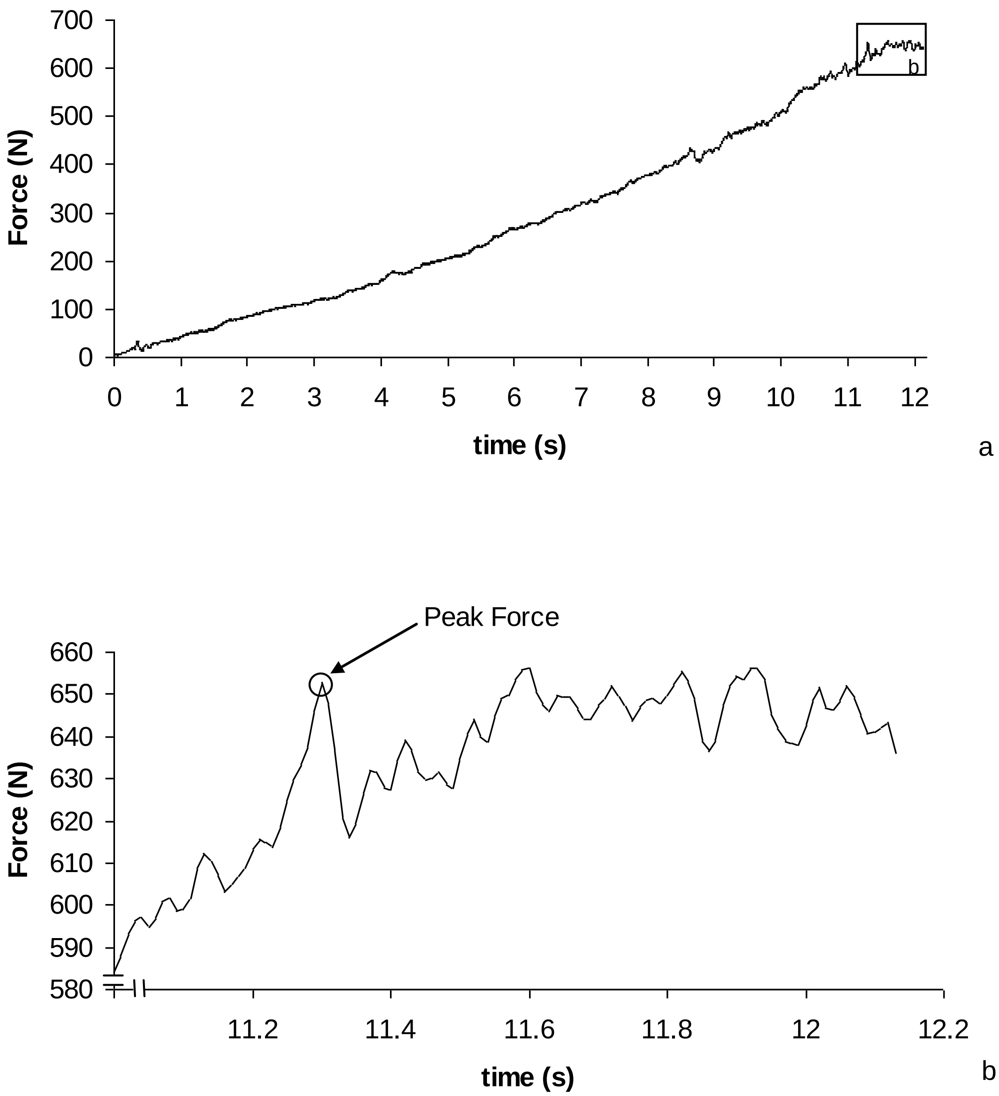


Figure 5.3.5 Example of a force-time curve (a). The expansion of the terminal portion of the curve (b) highlights the selection of the peak force.

There may be noise at peak efforts as there was likely a change in strategy such as breath holding, or other tricking movements that lead to an inability to maintain a maximal plateau as well as QF fatigue.

Once a trial for each of the d1, d2 and TL segment recordings was chosen, the trial with the greatest common PT force value was deemed the MVIC for that leg. By choosing the greatest common force, the ability to repeat such an effort was more likely. It also reduces the chances that any greater values were not a consequence of artifact or noise. Further to this, the MVIC measures a net extensor moment. The knee extensors may be able to generate greater force, were it not for the effect of hamstring activity that may be acting with a flexion moment. Hence, there may be potential reduction in knee extensor moment by not accounting for the contribution of hamstring activity. The focus was on measuring PT length at a common PT force across trials and not specifically the maximal possible PT force. Also by establishing a clinically applicable method, the possible reduction in knee extensor moment due to antagonist co-activation was not calculated.

The PT force recordings for d1, d2 and TL were truncated at the same common maximal PT force so the three recordings were ‘force matched’. This allowed values from 0 to 100% of the MVIC to be calculated for each force-time curve where PT force values were similar for recordings of d1, d2 and TL at 10% increments from 0 to 100% MVIC effort.

5.3.3.ii Ultrasound Imaging

5.3.3.ii - Patellar Tendon Cross-Sectional Area

Recorded digital video was converted to a series of bitmap (bmp) images utilising editing software (Apple, Final Cut Pro Version 5). The polygon selection on Image J software (NIH: http://rsb.info.nih.gov/nih-image/) was used to trace the border of the PT (Figure 5.2.2) (O'Brien, et al., 2010; Onambélé, et al., 2007a). The area within the tracing was measured to provide a value for the PT CSA recorded at the proximal attachment of the PT and this measure as used for the calculation of PT stress.

5.3.3.ii - Patellar Tendon Elongation

Once the d1, d2 and TL trial for each leg was determined the corresponding digital video recording of PT elongation was truncated (Apple, Final Cut Pro Version 5) to synchronise with the triggered pulse in the force data. The digital video was converted to a series of bit map (bmp) images at 25fps (Apple, Final Cut Pro Version 5). As force data and ultrasound imaging were time synchronised, the corresponding frame at 10% increments from 0 to 100% of the MVIC was identified. The image corresponding to each effort at 0, 10, 20, 30, 40, 50, 60, 70, 80, 90 and 100% MVIC effort had d1, d2 and TL measured manually from the respective images, utilising Image J software (http://rsb.info.nih.gov/ij) (Onambélé, et al., 2007a).

5.3.3.iii Calculation of In Vivo Patellar Tendon Mechanical Properties

The measures of PT CSA, elongation and force allowed calculation of mechanical properties listed in Table 5.3.1. The value of modulus is the quotient of stress by strain and takes the same units as stress as strain is a value with no unit.

Table 5.3.1 Patellar tendon mechanical properties calculated form measures of force and elongation.

Mechanical Property	Derived Variable Formula
Stiffness ($N.mm^{-1}$)	Δ Force/Δ PT length
Strain (%)	Δ Elongation / PT resting length
Stress (MPa)	Δ Force / PT CSA
Elastic Modulus (GPa)	Δ Stress/Δ PT strain

5.3.4 Derived variables

5.3.4.i Validity

The derived variables included in this part of the pilot investigation were d1, d2, D (where D = d1 + d2) and TL. Trials were force matched such that at 10% incremental increases in effort from 0 to 100% effort determined by PT force and the lengths of d1, d2, and TL were measured and recorded. Thus, values of d1, d2 and D were each compared to the corresponding value of TL from recordings during a single session to determine the validity of measuring PT elongation comparing the two methods; one of measuring the complete PT length and the second of measuring d1 and d2 independently and summing them to estimate of the total length.

5.2.4.ii Reliability

The derived variables for which reliability was established included strain, stiffness, stress and elastic modulus (Table 5.3.1). These were calculated using the measured values of PT length (D), PT force and PT CSA. To determine reliability of these derived variables, results from the two separate testing sessions were compared for each derived variable.

For calculation of stiffness and elastic modulus a common PT force was chosen at a near-maximal effort common to sessions 1 and 2 for each participant. A range of 200 N (from 300 N less the maximum to a value 100 N less the maximal value) was selected and d1 and d2 were measured and summed to provide D. Patellar tendon elongation was measured at this near-maximal force. The gradient of the force-length and stress-strain curves at this range of 200N of PT force were calculated to provide stiffness and elastic modulus respectively. Strain was calculated as the percentage change in D from rest (0% effort) to the near maximal effort (100N less the maximal effort). Stress was the near maximal effort (PT Force) divided by the PT CSA measured during that session. Thus, test re-test reliability was determined by comparing the derived variables between two testing sessions.

This was considered an accurate means to determine reliability as it incorporated error in all steps that may have led to an error in measure of the derived variable. This includes error in the measure of the PT moment arm, PT CSA, knee moment and PT elongation.

5.3.5 Data Analyses

5.3.5.i Validity

Linear regression analysis was performed to compare TL against d1, d2 and D by entering all measurements for PT elongation, for the 8 limbs into a spreadsheet (W.G. Hopkins, 2000) from one testing session. When significant linear correlations were identified, the mean and 95% confidence interval (CI) of the regression coefficient (slope) and Y intercept for elongation were determined.

Typical error values (SDdiff/$\sqrt{2}$, where SDdiff is the standard deviation of the difference scores) and 95% CI were determined for the paired comparisons of TL against D (W.G. Hopkins, 2000).

5.3.5.ii Reliability

Test-retest reliability was assessed by comparing the measure of PT force, elongation, strain, stiffness, CSA, stress and elastic modulus between sessions 1 and 2. Data were entered into a spreadsheet (W.G. Hopkins, 2000) with the intra-class correlation coefficient (ICC) and typical error values (95% CI) determined. The change in mean and 95% CI for between session measures were also calculated to identify any systematic differences.

5.4 Results

The PT elongation and strain values for the eight limbs tested are shown in Table 5.4.1 from 0% to 100% MVIC effort.

Table 5.4.1 Elongation (mm) and strain (%) values (mean SD)) for the 8 limbs (four participants) tested, measuring the total length, summed length (proximal plus distal), proximal segment and distal segment lengths.

	Total	Summed	Proximal	Distal
Elongation (mm)	3.4 (1.1)	3.4 (1.0)	1.7 (1.4)	1.7 (1.8)
Strain (%)	10.2 (3.8)	10.1 (3.9)	11.4 (12.2)	8.2 (10.0)

The force, elongation, stiffness, CSA, strain, stress, modulus and PT moment arm length measures for testing sessions 1 and 2 are shown in table 5.4.2.

Table 5.4.2 Patellar tendon force, elongation, strain, cross-sectional area, moment arm length, stiffness and elastic modulus values (mean (SD)) for testing sessions 1 and 2.

Variable	Session1	Session 2
Force (N)	2113.4 (460.3)	2108.9 (461.8)
Elongation (mm)	3.4 (1.0)	3.0 (1.0)
Strain (%)	10.2 (3.8)	9.9 (3.7)
Cross-Sectional Area (mm^{-2})	124.25 (18.8)	121.3 (20.3)
PT moment arm length (mm)	48.5 (2.5)	49.0 (1.8)
Stiffness ($N.mm^{-1}$)$_{200N}$	1873.7 (1016.7)	1602.5 (639.1)
Elastic Modulus (GPa)$_{200N}$	0.53 (0.29)	0.5 (0.2)

$_{200N}$ – Calculated as the gradient (force-length & stress-strain curves) at a near-maximal effort at a preset range of 200N for patellar tendon force

5.4.1 Validity

A total of 79 measurements, for the 8 limbs, were made for PT length measurements for each of total length (TL), proximal segment (d1) and distal segment (d2). When comparing the measures of PT length for TL against the summed length (D) where D = d1 + d2, Pearson's r = 1.0. For linear regression of TL against D the regression coefficient (slope) for elongation was mean 1.0 [95%CI] [0.99 to 1.00] and the intercept of the line of regression was 0 [-0.05 to 0.05] for elongation (Figure 5.4.1 (C)).

Correlation of TL against d1 or d2 demonstrated moderate to poor correlation with Pearson's r = 0.5 and r = 0.08 respectively (Figure 5.4.1 (A) and (B)). This shows that in isolation each segment only accounted for 51% and 49% of the TL length respectively.

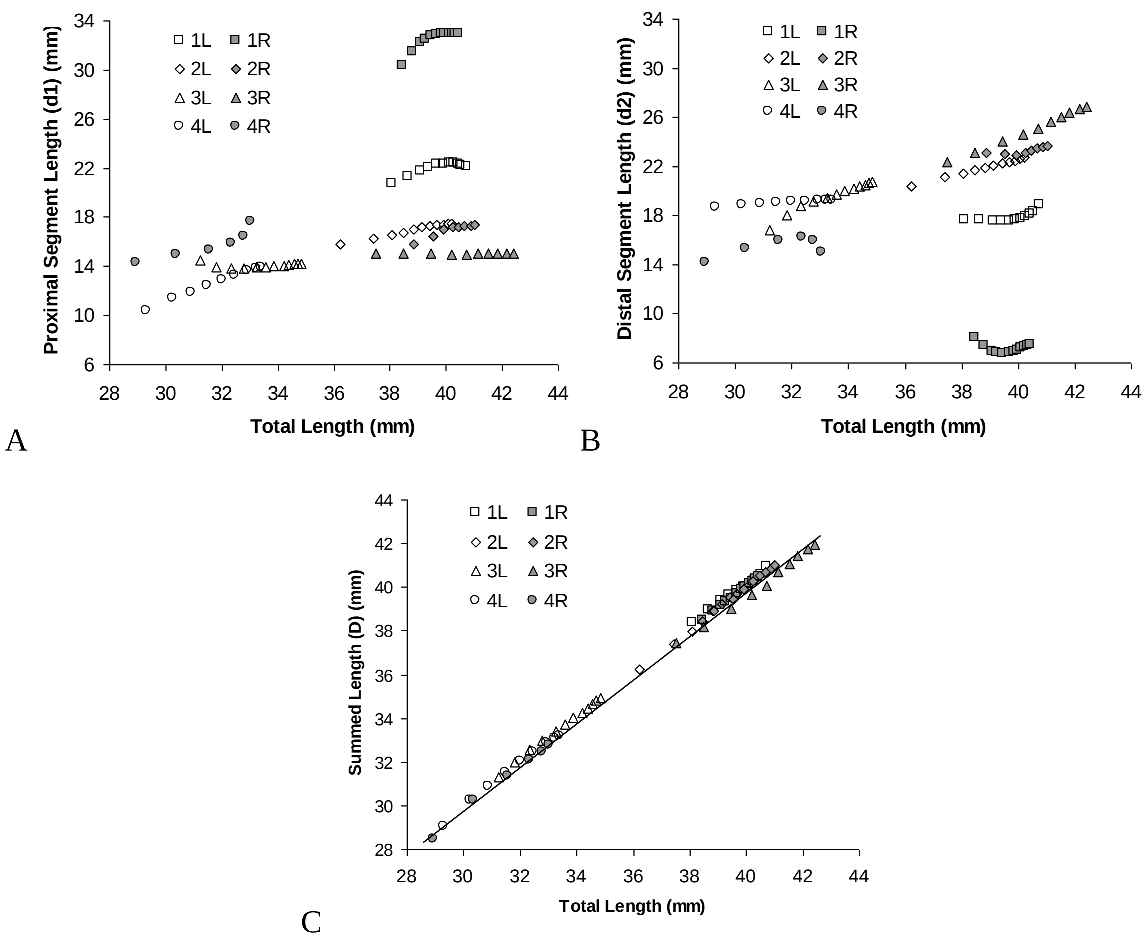


Figure 5.4.1 The association between the measure of total patellar tendon (PT) length (TL) during progressively increasing isometric load against the proximal (d1) (A), distal (d2) (B) and summed (D) lengths (C) are shown for left (L) and right (R) legs for the 4 participants. There is poor correlation of TL against either d1 or d2 in isolation. By summing d1 and d2 the measurement error is corrected resulting in a highly linear (solid black line in C) and concordant estimate of total PT length. The findings show that using only a proximal or distal method of estimating elongation the data is not valid.

5.4.2 Reliability

Test re-test reliability of measured PT moment arm length, force, elongation stiffness, strain and CSA was very high and for elastic modulus was high (Table 5.4.3).

Table 5.4.3 Between session test re-test reliability (mean (95%CI)) for derived variables.

Derived Variable	N	ICC	Typical Error	Change in mean
Force (N)	8	1.00	5.47 (3.62 - 11.14)	-4.58 (-11.05 – 1.89)
Elongation (mm)	8	0.99	0.1 (0.1 – 0.5)	-0.1 (-0.17 - -0.00)
Strain (%)	8	0.99	0.33 (0.21 - 0.66)	-0.3 (-0.69 – 0.08)
CSA (mm^2)	8	0.86	7.45 (4.93 – 15.17)	-2.96 (-11.78 – 5.85)
PT moment arm length (mm)	8	0.99	0.53 (0.35 – 1.09)	0.00 (-0.63 – 0.63)
Stiffness $(N.mm^{-1})_{200N}$	8	0.84	417.71 (276.18 – 850.15)	-271.21 (-765.07 – 222.65)
Stress ($N.mm^{-2}$)	8	0.96	1.14 (0.75 – 2.31)	0.34 (-1.00 – 1.69)
Elastic Modulus $(GPa)_{200N}$	8	0.74	0.14 (0.09 - 0.28)	-0.06 (-0.22 - 0.1)

$_{200N}$ - Calculated at a range of 200 N (from 300 to 100N below maximal effort)

5.5 Discussion

This pilot study demonstrated that the use of an ultrasound shadow frame of reference across the PT was able to provide an excellent estimate of total PT length when proximal and distal segment lengths were summed. In isolation, each segment length correlated poorly with PT length. This may be explained by the shift in the frame of reference between the PT bony attachment (inferior pole of patella and tibial tuberosity) and the marker shadow when observing proximal or distal segment only or movement of the transducer relative to the skin. An error in one segment will reflect an equal and opposite error in the other segment. Previous studies have demonstrated that PT elongation was underestimated when distal (tibia) displacement was not accounted for (P. Hansen, et al., 2006; Onambélé, et al., 2007a). The main feature in this study was that the measure of PT elongation measuring proximal and distal segment length was highly accurate compared to a gold standard reference of measuring the complete PT length rather than comparing the PT mechanical properties (strain, stiffness and modulus) from measuring proximal and distal segment lengths only.

There was a strong relationship (linear or curvilinear) between the measure of TL and d1 plus d2 for each individual leg (Figure 5.4.1, A and B) and a high degree of variability between legs and individuals. Thus, the error associated between a measure of PT total length and segment length was a consequence of the experimental set-up rather than random error and was predictive of incremental elongation for the PT segment length.

Although the reliability of measuring the proximal PT segment length was previously shown to be high (Burgess, et al., 2009c; S. J. Pearson, et al., 2006; S. P. Pearson, et al., 2007; N. D. Reeves, et al., 2003a), this does not validate the use of a measure of PT segment length as being transferable to other legs or testing sessions. The result of summing proximal and distal values, reflective of total PT length, will control for the systematic error associated with experimental set-up. There was a strong linear relationship between the measure of total length and summed length, for all individuals and legs (Figure 5.4.1, C). This pilot study has shown that by summing d1 and d2 segment lengths at a common force, the error associated with experimental set-up was cancelled.

In the current study, compared to the summed length (D), PT strain was overestimated by 12.8% when measured using the proximal segment length change and conversely it was underestimated by 19.2% when assessed using the distal segment length only. In this current study the distal segment length techniques reduced the estimate of PT strain. This is in contrast to the two previous studies that reported, accounting for distal attachment (tibial) motion increased the estimate of PT strain (P. Hansen, et al., 2006; Onambélé, et al., 2007a). In the current study, one reason for this was the fact that there was a large variation between individuals and legs therefore the mean error (under or over estimation) may have been related to the differences between participants.

In the current study, validity of recording trials for the proximal and distal segment lengths for determination of PT elongation was demonstrated in a population where the total PT length was able to be visualised. These findings show that it was possible to determine PT elongation using this technique and that it was representative of PT elongation when the total PT length may not be visualised and was measured as the sum of proximal and distal segments. This technique will be specific to populations such as taller athletes that participate in volleyball or basketball, for measuring PT elongation and determining mechanical properties.

The values for PT strain at 100% effort (10.1%) lie between the values for those reported previously that range from ~6.8% (P. Hansen, et al., 2006) and ~16.3% (Onambélé, et al., 2007a) and approximate values for adults reported in other studies (Maganaris, et al., 2004; O'Brien, et al., 2010; S. P. Pearson, et al., 2007; N. D. Reeves, et al., 2003a). Differences in the results may be due to differences in the sample populations although strain has not been shown to be different between genders (O'Brien, et al., 2010; Onambélé, et al., 2007a) with the results of the current study similar to those for the elderly (N. D. Reeves, et al., 2003a). At very low load knee moments (10Nm), *in vivo* cine MRI estimated PT strain to be 6.6% with a maximum of 11% (Sheehan & Drace, 2000). The values of PT strain in the current study are supportive of the previous literature considering the differences in methods using either ultrasound or MRI.

Methods of measurement were similar between studies with the use of B mode ultrasound, dynamometry and PT elongation being tested in 90^{o} knee flexion. Differences in strain may be attributed to the speed of knee extensor contraction that was maintained at 10s for 3 participants and 5s for one participant in the current study. Pearson et al. (2007) demonstrated that there was greater strain at a slower (10-12s) contraction speed of mean 8.0% (SEM) (0.9) compared to faster (3-4s) contraction speed with a strain of 4.5% (0.5) (S. P. Pearson, et al., 2007). Thus, maintaining a within-subject design for validation and reliability measures controlled for any differences due to contraction speed.

The method for measure of PT elongation may also lead to differences in values for PT mechanical properties. Care was taken in this pilot study to measure the PT length at the posterior portion of the PT that had easily identifiable bony attachments since previous studies have demonstrated differential strain between anterior and posterior portions of the PT (Almekinders, et al., 2002; Basso, et al., 2002) and variance in strain between medial, central and lateral one third portions of the PT from ~10 to ~12% (DeFrate, Nha, Papannagari, Moses, Gill, & Li, 2007). Furthermore, PT strain has been calculated from measures of bone displacement (patella and tibia) (P. Hansen, et al., 2006; Onambélé, et al., 2007a) and may not be specific to either the anterior or posterior portion of the PT. Kongsgaard et al. (2007) maintained a full view of the PT length from origin to insertion during ramped MVIC knee extensor contractions, however, the ultrasound recording illustrates the difficulty in identifying a bony attachment that is consistently matched across frames as force increases (Kongsgaard, et al., 2007). By not visualising the complete tendon length, there may be an overestimation of the PT length if the tibia moves proximally and underestimation if the tibia moves distally.

In the current procedure for measure of PT length at a given frame, care was taken to scroll the images back and forth from the frame of interest (pre-determined from matching a frame number to a specific force value). This entailed identification of the posterior portion of the PT to the bony attachment of the patella proximally and tibia distally (Figure 5.5.1). A clear identification of the 'step' shape of the tibial attachment of the PT allowed an easily identifiable bony landmark common to all frames at different PT forces. This method of identification of bony landmarks is similarly illustrated in a recent study (O'Brien, et al., 2010). As the skin marker was adhered to the skin and common to both sets of images, movement of the shadow from the skin marker would be common when recording either the proximal or distal segment.

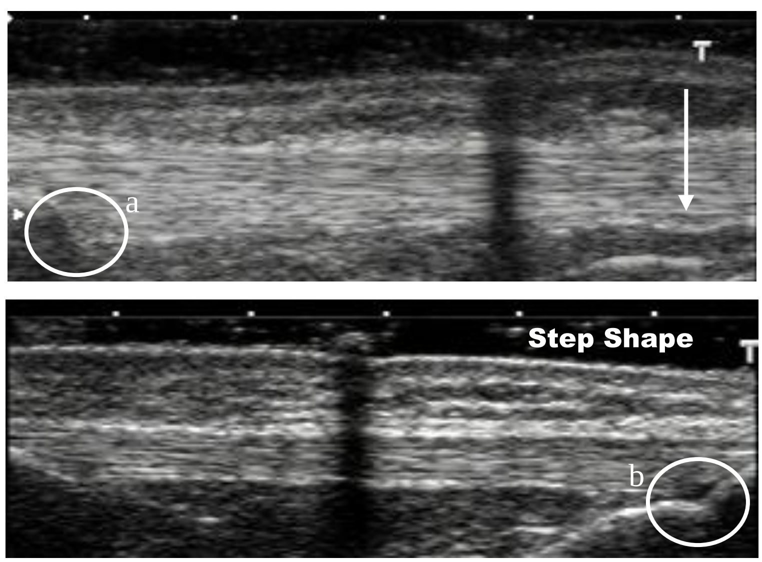


Figure 5.5.1 Proximal attachment of the posterior portion of the patellar tendon (PT) to the patella (white circle) (a). Distal attachment of the posterior portion of the patellar tendon to the tibia (white circle) (b). Note that in a, there is no distinct site of PT distal attachment to the tibia (white arrow) compared to (b) where there is a 'step' shape to the tibial tuberosity site of attachment of the posterior portion of the PT.

Investigation of total PT strain has been assumed to be proportional to the proximal segment (Kubo, et al., 2005a; S. J. Pearson, et al., 2006; N. D. Reeves, Narici, & Maganaris, 2003b). The current study suggests there is a poor correlation between proximal segment strain and total tendon strain with the proximal strain overestimating the total PT length strain. This is in contrast to an earlier study where it was observed that proximal strain underestimated PT strain (Onambélé, et al., 2007a) with a change in proximal length divided by total PT resting length. In the current investigation the denominator was the resting length of the proximal segment and will yield a greater segmental strain value, possibly overestimating total PT strain. Using an external frame of reference does not allow measure of a specific segment's strain contribution to total strain as there may be relative movement of the PT and echo-absorptive marker.

Ultrasound imaging of the PT is unique to other tendinous structures in the human body with bony attachments proximally and distally. Typically this type of tissue is classified as a 'ligament', however considering the PT is responsible for transmission of force from the QF, the PT is susceptible to the same forces as the contractile component. Direct geometric measure of the PT is made easier due to identification of the limits of the tendon; however commonly available ultrasound technology does not always allow visualisation of these landmarks on a single image. Therefore, the selection of a skeletally mature sample population of volunteers that allowed complete PT visualisation of the PT was deemed necessary. This allowed a geometric measure of the PT at the posterior portion to be made in full visualisation and of each segment (proximal and distal) when the tendon deformation of the corresponding segment was unknown. Furthermore, this study focused on a direct measurement of the posterior portion of the PT rather than from a point on the patella to a point on the tibia that may be attached to different portions of the PT. Considering the variability in strain between anterior and posterior bundles of the PT, this study establishes a specific method for measurement of the posterior portion of the PT.

Although the measure of knee moment using dynamometry is widespread, finding a reliable method for accurately measuring the PT moment arm length for calculation of PT force was challenging. The aim was to develop a measure of PT force in the clinical setting that was inexpensive and easy-to-use. The caliper technique was developed as it would be used in future studies for assessing PT mechanical properties in a larger sample population of athletes in a training and clinical setting. As such it was important to avoid high costs and allow measure of the PT moment arm quickly with minimal risk and time constraint to participants.

The range of values measured for PT moment arm length in the current study (44 to 57 mm) is similar to values reported by others (39.9 to 60 mm) (Baltzopoulos, 1995; Herzog, et al., 1993; Kellis, et al., 1999b; Krevolin, et al., 2004; Maganaris, 2004; Tsaopoulos, et al., 2006a; Wretenberg, et al., 1996). The current method maintained the transepicondylar axis as the knee joint axis of rotation (Asano, et al., 2005; Krevolin, et al., 2004; Yoshino, et al., 2001) in contrast to previous studies that have defined the knee joint axis of rotation as the tibia-femur contact point (Kellis, et al., 1999b; Maganaris, 2004; Maganaris, et al., 2005; N. D. Reeves, et al., 2003a; Wretenberg, et al., 1996) or cruciate ligament cross-point (Tsaopoulos, et al., 2006a). A number of studies have used the methods listed to estimate PT moment arm length (Maganaris, et al., 2005; Maganaris & Paul, 1999; O'Brien, et al., 2010; N. D. Reeves, et al., 2003a; N. D. Reeves, et al., 2003b) or averages of values reported in these studies (Burgess, et al., 2009a; Burgess, et al., 2009b; Onambélé, et al., 2007a; Onambélé & Pearson, 2007b). Similarly, PT force estimates in some studies (Couppé, et al., 2008; P. Hansen, et al., 2006; Kongsgaard, et al., 2007; Kubo, et al., 2006) were based on PT moment arm length estimates used from estimation techniques based on the findings of 6 cadaveric limbs that determined the moment arm length of the vastus lateralis, vastus medialis, vastus intermedius, rectus femoris and biceps femoris (Visser, Hoogkamer, Bobbert, & Huijing, 1990). Using previously reported PT moment arm length values that are taken based on a different sample population may lead to an inaccurate estimate of PT force, as the PT moment arm length estimate is not specific to the participant for whom knee moment is measured.

The caliper technique introduced in the current study was a direct geometric measure of the PT moment arm length whereby the line of pull of the PT was perpendicular to the PT moment arm. It was shown to be reliable, easy-to-use and inexpensive. These issues were of importance in the population of elite level athletes being tested in subsequent studies as they were generally taller and more muscular than the general population. The number of participants being tested in the subsequent studies was also relatively greater than previous studies. The current method of measuring PT moment arm length was appropriate and had values for PT moment arm length in concordance with previous reports (Tsaopoulos, et al., 2006a), providing a subject-specific estimate, as there may exist variability in the PT moment arm length based on inter-subject variability in physiology (Maganaris, et al., 2005). In maintaining the PT moment arm length perpendicular to the line of pull of the patella, the knee was flexed from 90° to 110° and approximates the knee angle (90º knee flexion) during recording of PT elongation and knee moment. Although the caliper technique does not account for the distance from the skin to the bony site of the attachment of the posterior portion of the PT to the inferior pole of the patella, this is relatively consistent between individuals as the inferior pole is easily palpable in 90° knee flexion. Future studies may consider validating the described caliper method to imaging techniques.

5.5.1 Strengths and Limitations

Strengths of this investigation are that as complete PT length was measured there was no need to correct for knee rotation that may contribute to tendon elongation, as proximal and distal attachments (P. Hansen, et al., 2006; Onambélé, et al., 2007a) are accounted for. The typical errors associated with distal displacement of the tibia were reported as ranging from 3.7% to 9.6% for strain, stiffness and modulus (P. Hansen, et al., 2006). Hansen et al. also demonstrated high within-day reproducibility in support of the current findings for PT strain, stiffness and modulus (P. Hansen, et al., 2006). Furthermore, maintaining sessions 30mins apart demonstrated reliability to be excellent as it has been reported that there PT mechanical properties may change due to the effect of time-of-day (Onambélé, et al., 2007b) or activity (Kongsgaard, et al., 2007; Kubo, Akima, Ushiyama, Tabata, Fukuoka, Kanehisa, & Fukunaga, 2004; Kubo, et al., 2006).

A limitation in measuring PT length using ultrasound was that only a single, sagittal plane of tissue was viewed. As tensile force was applied the plane of tissue being viewed may change as there was movement between the skin and PT (Sheehan, Zajac, & Drace, 1999). Another source of error was that created by moving the transducer from one segment to the other with successive MVIC efforts. Systematic error was reduced in this study with a randomisation of the order of testing either proximal, distal, and total lengths.

A limitation in this study was not correcting knee moment for the contribution of load due to antagonist (hamstring) muscle co-contraction. The method of using EMG to accurately quantify the contribution of the load of the hamstring, although previously used (Bojsen-Møller, et al., 2003; Burgess, et al., 2009c; S. J. Pearson, et al., 2005; N. R. Reeves, et al., 2009; Westh, et al., 2008) is problematic and this is endemic in the literature (Grabiner, Koh, & Miller, 1992; Grabiner, R., Hawthorne, & Hawkins, 1989; Kellis, 2003; Kellis & Baltzopoulos, 1999a; Kellis & Katis, 2008; Kellis, Kouvelioti, & Ioakimidis, 2005; Kingma, Aalbersberg, & van Dieën, 2004; Krishnan & Williams, 2010). The use of specific EMG recorded signal characteristics is not reliable nor a good solution to predict the level of force resulting from muscle contraction.

Specifically, in the testing protocol used, variance in the estimate of knee moment due to antagonist muscle activity as determined by EMG may occur due to differences in EMG signal between the muscle bellies of the hamstring muscle group (Aagaard, Simonsen, Andersen, Magnusson, Bojsen-Møller, & Dyhre-Poulsen, 2000), differences in EMG signal between trials due to differences in the speed of the maximal isometric contraction, a lack of a gold standard in EMG normalisation procedure (Basmajian & DeLuca, 1985; Kellis, 1998; Kellis & Baltzopoulos, 1996; Knutson, Soderberg, Ballantyne, & Clark, 1994) and EMG signal cross-talk between different muscle bellies and muscle groups (Etnyre & Abraham, 1988; Grabiner, et al., 1992; Kellis, 1998).

Further problems relate to the interpretation of antagonist hamstring EMG activity during isometric knee extension. That is the ability to to determine the hamstring contribution to knee moment. One of these factors is the difference in moment arm length. The moment arm length of the line of the pull of the hamstring tendons is 50 to 80% of the patellar tendon moment arm length (Ellis, Seedom, Amis, Dowson, & Wright, 1979). Therefore, even if it is assumed that a linear increase in EMG signal is proportional to a linear increase in force, the net contribution to knee moment is different between the hamstring and quadriceps muscle groups. Furthermore, differences in the physiological cross-sectional area (PCSA) of the agonist and antagonist groups will lead to misinterpretation of EMG signal. It is accepted that muscle force is proportional to PCSA (Ellis, et al., 1979; K. R. Kaufman, K. N. An, W.

Litchy, J., & E. Y. Chao, 1991; K. R. Kaufman, K. N. An, W. J. Litchy, & E. Y. Chao, 1991; Roy & Edgerton, 1992; Wickiewicz, Roy, Powell, & Edgerton, 1983); therefore, if EMG activity is the same for both agonist and antagonist muscles, then force will not be comparable.

The method of determining antagonist (hamstring) force contribution to knee extension moment includes a number of possible sources of error. Collecting EMG signal from a single muscle only, such as biceps femoris (Burgess, et al., 2009c) may not represent equal force contribution across all muscles that flex the knee. Differences in the EMG signal due to different muscle action, whereby the hamstring will act to shorten when acting as an agonist and the same muscle will lengthen when acting as an antagonist. Therefore, there may be a difference in the moment or force between the two muscle actions (Kellis, 1998). There is also a difference in electromechanical delay between different muscle groups (Cavanagh & Komi, 1979). Although the movement tested is considered isometric, there is some compression of soft tissue and foam of the transducer pad during isometric knee extension and flexion. The set up in a dynamometer does not fully prevent the knee maintaining a fixed joint angle. This will affect the collected signal between agonist and antagonist muscle groups, especially when the same muscle group has EMG activity recorded as either an agonist or antagonist. This difference in EMG activity between muscle actions will therefore be a source of error in estimating force when the hamstring acts as an antagonist and EMG activity is calculated with reference to force recorded during a maximal agonist effort. The use of a dynamometer during a ramped MVIC of the quadriceps femoris was used and measured the "resultant (net) joint moment or force around the knee joint" (Kellis, 1998). The ability to quantify the contribution of the antagonist muscle group remains problematic via EMG recording and was not used due to the multiple sources of error that may contribute to the estimate of hamstring force during knee extension.

Patellar tendon stiffness and elastic modulus were calculated from the gradient of the force elongation curve and stress strain curves respectively, from 70 to 80%MVIC effort (chapter 6). The candidate has supported the use of this method (pg. 196) as patellar tendon elongation was either not measured or less than the typical error (chapter 5). Therefore, while the proportion of contribution to patellar tendon elongation due to quadriceps and hamstrings activity may not be known, it was accepted that at the near maximal efforts (70 to 80% MVIC) the ‘net’ knee extension force (quadriceps femoris and hamstring activity) was responsible for the ‘net’ patellar tendon elongation. It was not the aim of these studies to determine how much of the patellar tendon elongation femoris was due to the activity of the quadriceps only. The data on knee moment was reliable and consistent with previous studies (Burgess, et al., 2009b; S. J. Pearson, et al., 2005; S. P. Pearson, et al., 2007). While it may be accepted that co-activation affects the data, this may be systematically different

5.5.2 Summary and Conclusion

This study has demonstrated that the sum of the proximal and distal segment elongation when assessed on two different occasions was highly concordant with the total length changes in the PT under a large range of load. By demonstrating this, it is clear that methods utilising an approach to measure the proximal or distal segment only, are not valid.

By overcoming the technological challenge of not being able to visualise the complete PT length using ultrasound, measurement of the PT mechanical properties in taller individuals may be made. This was of specific interest in this thesis that investigated elite jumping athletes, who are susceptible to patellar tendon changes and tendinopathy. This study defines a method for measuring elongation of the posterior portion of the PT and derived variables specific to the proximal attachment that is the most common site of patellar tendinosis.

6 Chapter Six: Patellar Tendon Mechanical Properties in Elite Athletes

6.1 Chapter Aims

This chapter aims to:

- compare patellar tendon mechanical properties between groups of elite level jumping and non-jumping athletes.
- compare patellar tendon mechanical properties between genders in a jumping sport.
- compare patellar tendon mechanical properties between the left and right sides of elite level jumping and non-jumping athletes.
- compare patellar tendon mechanical properties between athletes based on dominance.
- compare mechanical properties between normal and pathological patellar tendons.

6.2 Introduction

Jumping athletes use loading strategies during sport that result in relatively high forces at the knee when landing (chapter 3) in sports such as volleyball (Bisseling, et al., 2008; Marquez, et al., 2009; D. P. Richards, et al., 1996; Sheppard, et al., 2009) and basketball (Louw, et al., 2006). Jumping and landing activities in these sports place high load on the PT that may be implicated in the pathogenesis of patellar tendinopathy (Elvin, et al., 2009). Not surprisingly athletes participating in volleyball and basketball are reported to have a high prevalence of patellar tendinopathy (Cook, et al., 1997; Cook, et al., 1998; Ferretti, 1986; Lian, et al., 2005; Witvrouw, et al., 2001).

The predominant location of patellar tendinopathy is at the bony attachments. Patellar tendinopathy is more common at the proximal attachment (at the PT insertion to the inferior pole of the patella) than distally where the PT is attached to the tibial tuberosity (Pfirrmann, et al., 2008; Sarimo, Sarin, Orava, Heikkilä, Rantanen, Paavola, & Raatikainen, 2007). A number of studies have demonstrated changes on ultrasound imaging in asymptomatic jumping athletes including hypoechoic areas and tendon thickening that are associated with patellar tendinopathy (Cook, et al., 2001a; Cook, et al., 2000b; Cook, Kiss, Khan, Purdam, &

Webster, 2004b; Gemignani, et al., 2008; Gisslèn, Gyulai, Söderman, & Alfredson, 2005; Peace, Lee, & Healy, 2006).

A number of factors may affect PT load. Each sport may expose the PT to a different magnitude of load. There are also differences in lower limb loading strategies between men and women (Fagenbaum & Darling, 2003; Gehring, Melnyk, & Gollhofer, 2009; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Padua, et al., 2005; Urabe, Kobayashi, Sumida, Tanaka, Yoshida, Nishiwaki, Tsutsumi, & Ochi, 2005) that may contribute to different PT loads. Patellar tendinopathy may also affect the ability of the PT to withstand load (Cook, Khan, Kiss, Purdam, & Griffiths, 2001b; Khan, Maffulli, Coleman, Cook, & Taunton, 1998 ; Kongsgaard, et al., 2009; Kujala, Osterman, Kvist, Aalto, & Friberg, 1986; Peers, et al., 2005). This is of interest as ultrasound imaging of the PT in individuals with patellar tendinopathy shows changes including tendon thickening and hypoechoic areas (Cook, et al., 1998; Cook, et al., 2001a; Cook, Khan, Kiss, & Griffiths, 2000a; Cook, et al., 2000b; Cook & Malliaras, 2005; Grassi, Filippucci, Farina, & Cervini, 2000; Peace, et al., 2006). Although imaging by itself cannot classify tendon health, as tissue changes on ultrasound imaging may be present in the absence of pain (Cook, et al., 2001a; Cook, et al., 2005), the PT mechanical properties of patellar tendons with tendinopathy in elite jumping and non-jumping athletes is not known.

A number of other factors may affect the PT. Cross-sectional studies evaluating PT mechanical properties have demonstrated differences based on gender with greater tendon stiffness in men than women (Onambélé, et al., 2007a; Westh, et al., 2008). In contrast, Kongsgaard et al. (2009) reported no significant differences in PT mechanical properties between normal tendons and those with patellar tendinopathy in non-elite athletes (Kongsgaard, et al., 2009). Inter-side knee extensor strength asymmetry also affects the PT where the stronger side had greater PT stiffness and elastic modulus (Couppé, et al., 2008). Similarly, PT donor legs for anterior cruciate ligament reconstruction had lower modulus and greater PT CSA than the contralateral side (N. R. Reeves, et al., 2009). These factors and those of exposure to training as discussed earlier, suggest that the PT can adapt to loading.

Apart from the type of sport participation, the pattern of loading of the lower limb during landing may influence the specific loading profile of the PT (Reilly, et al., 2009; Sheppard, et al., 2009). Furthermore, PT loading may be affected by the task being assessed and whether there is a single-leg loading strategy (Louw, et al., 2006; Marquez, et al., 2009; D. P. Richards, et al., 2002; D. P. Richards, et al., 1996; J. G. Richards, et al., 2008) or double-leg loading strategy (Schot, Bates, & Dufek, 1994; Tillman, et al., 2004). Laterality asymmetry and dominance may also impact on the lower limb loading strategy. For example, greater knee flexion angle on landing was associated with an increased risk of developing patellar tendinopathy on the left side (D. P. Richards, et al., 1996). During landing from a spike jump in volleyball there is an inherent asymmetrical strategy that results in a step-close landing technique (Marquez, et al., 2009; Tillman, et al., 2004). This is consistent with the centre of pressure being maintained closer to the left than right sides (Marquez, et al., 2009; Tillman, et al., 2004). Double-leg landing strategy has also been shown to have variability between sides especially in relation to joint moment at the knee (Schot, et al., 1994). In summary, elite athletes are exposed to asymmetrical loading of the lower limbs due to the nature of their sports as well as the preference of their natural laterality or dominance (Beling, Wolfe, Allen, & Boyle, 1998; Gabbard & Hart, 1996a). To date no studies have examined the impact of laterality on the mechanical properties of the PT in a homogenous elite athletic population.

Loading behaviour may lead to PT hypertrophy as an increase in PT CSA has been reported following knee extensor strengthening exercise (Kongsgaard, et al., 2007; Seynnes, Erskine, Maganaris, Longo, Simoneau, Grosset, & Narici, 2009). Interestingly, PT CSA was unchanged following unloading (23 days of limb suspension), however, stiffness and elastic modulus, were reduced (de Boer, Maganaris, Seynnes, Rennie, & Narici, 2007). Whether PT CSA is different between elite level athletes participating in jumping and non-jumping sports is not known. As different sports may expose the PT to different lower limb loading behaviour, evaluation of differences in PT CSA based on the type of sport may distinguish an athlete’s susceptibility to tendon hypertrophy or tendinopathy.

The purposes of this chapter were to measure *in vivo* PT mechanical properties in elite jumping and non-jumping athletes and examine the force-elongation and stress-strain relationships.

6.2.1 Hypotheses

The following null hypotheses were tested:

1. That there is no difference in PT mechanical properties between sporting groups.
2. That there is no difference in PT mechanical properties between men and women participating in the same sport.
3. That there is no difference in PT mechanical properties between normal left and right patellar tendons.
4. That there is no difference in PT mechanical properties within each sporting group when stratified by side based on dominance.
5. That there is no difference in mechanical properties between normal and pathological patellar tendons.

6.3 Methods

6.3.1 Study Design

This was a cross-sectional study design. All participants underwent a clinical examination and had PT mechanical properties measured.

6.3.2 Participants

Sixty athletes from the Australian Institute of Sport athlete population volunteered to participate (Table 6.3.1). Participants were full-time athletes from jumping (volleyball and basketball) and non-jumping (swimming) sports and undertook approximately 30 hours/week of training and competition.

Table 6.3.1 Descriptive characteristics of the athletes (mean (SD)).

Sport Group	Number	Age (years)	Height (m)	Weight (kg)
Men's Volleyball	23	18.91 (3.18)	2.00 (0.06)	91.39 (9.04)
Men's Basketball	17	17.00 (1.00)	1.96 (0.11)	88.22 (12.94)
Women's Basketball	14	17.00 (0.78)	1.83 (0.08)	78.07 (0.60)
Male swimmers	4	20.25 (1.29)	1.83 (0.13)	76.35 (17.42)
Female swimmers	2	20.00 (0.71)	1.73 (0.15)	64.00 (9.05)

Participants were excluded if they had a current history of pain that may have been aggravated by the testing procedure or if they reported any known medical or physical condition that prohibited them from physical exertion, activity or any condition that may be aggravated by the testing procedure(s) (eg. epilepsy, anterior cruciate ligament reconstructive surgery, sero-negative disorders, etc.). Each participant self-reported their sport, player position, height (nearest centimetre) and body mass (nearest kg) as these measures were taken weekly as part of their training program.

This investigation was approved by the La Trobe University Faculty Human Research Ethics Committee (Appendix A) and the Australian Institute of Sport Research Ethics Committee (Appendix B). All participants provided signed informed consent (Appendix C). If participants were under 18 years of age, written parental/guardian consent was obtained. All participants were able to withdraw their consent at any time during testing.

6.3.3 Limb Dominance

Leg and arm dominance in activities of daily living (ADL) (Chapman & Chapman, 1987; Karev, 2009; Peters, Reimers, & Manning, 2006; Searleman & Porac, 2003) and in sport-specific tasks were determined. The participant reported their preferred side in the following tasks: hand writing, kicking leg, jumping leg, landing leg, shooting (basketball) or spiking (volleyball) hand (Ford, Myer, & Hewett, 2003).

6.3.4 Clinical Examination

A clinical examination was performed prior to testing. All clinical examinations were performed by a qualified and experienced sports physiotherapist who was familiar with a clinical diagnosis of patellar tendinopathy (Hyman, 2008; Khan, et al., 1998 ; Peers, et al., 2005). The sports physiotherapist was independent of the examiner who was measuring PT mechanical properties. Clinical examination recorded pain (body chart and visual analogue scales) (Strong, Sturgess, Unruh, & Vincenzino, 2002) and function (Victorian Institute of Sport Assessment (VISA) questionnaire) (Visentini, Khan, Cook, Kiss, Harcourt, & Wark, 1998).

6.3.4.i Body Chart

Each participant was asked about pain in the left or right knees. If knee pain was reported by the participant, the area of pain was recorded on a knee chart by the examiner (Maitland, 1975; Strong, et al., 2002) (Figure 6.3.1).

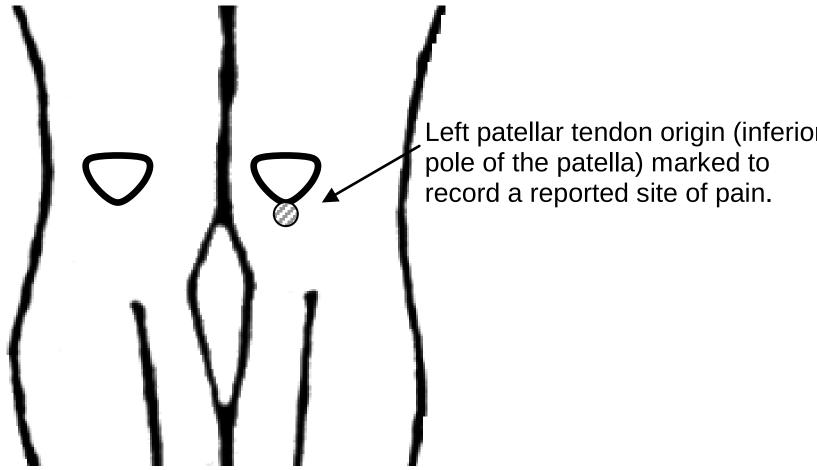


Figure 6.3.1 Knee chart with the left patellar tendon origin marked (grey shaded circle) indicating an area of reported pain.

6.3.4.ii Visual Analogue Scale – Pain Intensity

The average intensity of pain experienced in the past week during training and competition was recorded (Strong, et al., 2002) on a visual analogue scale (VAS). The VAS was a 100 mm horizontal line where the left end represented no pain and the right end represented the worst experience of pain (Strong, et al., 2002). A VAS was completed for the left and right sides.

A single leg decline squat test was used to assess PT pain response (Cook, et al., 2000a; Cook, et al., 2005; Purdam, et al., 2003). Each participant performed a single leg decline squat to 90º knee flexion on a decline board (23º) on the left and right legs. The participant recorded pain intensity on a VAS chart during this test.

6.3.4.iii Victorian Institute of Sport Assessment Questionnaire

The VISA has been shown to be reliable and valid in quantifying dysfunction due to patellar tendinopathy (Appendix M) (Frohm, et al., 2007a; Robinson, Cook, Purdam, Visentini, Ross, Maffulli, Taunton, & Khan, 2001; Visentini, et al., 1998; Young, et al., 2005) and was completed by each participant for each leg.

6.3.5 Patellar Tendon Mechanical Properties

Methods to determine PT mechanical properties (strain, stiffness and elastic modulus), force and CSA were described in chapter 5. The total PT length from bony origin to insertion during a MVIC was unable to be visualised in this study because the study cohort was taller and the total length of the PT did not lie within the window of the ultrasound transducer viewing window. Therefore, recordings of proximal and distal segments were made. Furthermore, PT CSA was measured at three sites along the site of the PT (proximal, mid and distal) (Burgess, et al., 2009b; Maganaris, et al., 2004; O'Brien, et al., 2010; Onambélé, et al., 2007a; Onambélé, et al., 2007b).

6.3.6 Data Processing

<u>6.3.6.i Clinical Examination</u>

Clinical examination classified the left and right PT of each participant as normal, imaging tendinopathy, clinical tendinopathy or clinical and imaging tendinopathy. The clinical classification for each is defined below and is consistent with the previously used classification of PT health (Cook, et al., 2001a; Hyman, 2008; Khan, et al., 1998 ; Malliarias & Cook, 2006; Peace, et al., 2006; Peers, et al., 2005).

Clinical tendinopathy required that the area of pain was located at the inferior pole of the patella (Peers, et al., 2005). The classification also included one or more of the following findings:

- Pain at the inferior pole of the patella during the last one week of activity
- Pain at the inferior pole of the patella during single leg decline squat.
- Reduced VISA questionnaire score (<80).
- Normal appearance of the PT on ultrasound imaging.

Imaging tendinopathy was assessed on a sagittal ultrasound image recorded during *in vivo* ultrasound assessment of PT mechanical properties by an independent and experienced sports physician (chapter 5, section 5.3). Imaging tendinopathy was defined by the presence of a hypoechoic region and/or thickening of the proximal tendon attachment (Figure 6.3.2 a and

b) (Cook, et al., 1998; Cook, et al., 2001a; Hoksrud, et al., 2008; Peace, et al., 2006; Peers, et al., 2005). The ultrasound was assessed by the sports physician who was blind to the findings of the clinical examination.

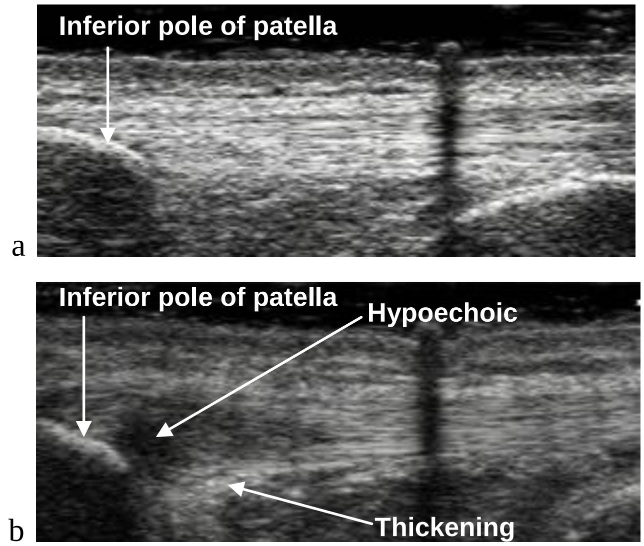


Figure 6.3.2 (a) Sagittal ultrasound image of the proximal segment of the patellar tendon classified as normal (a) and imaging tendinopathy (b) that has a hypoechoic area and thickening at the proximal attachment to the inferior pole of the patella.

A participant with a normal PT reported no pain at the site of the PT, no report of pain in the last week of activity, no pain on single leg decline squat, no reduction in the VISA score (> 80) and no imaging abnormality on ultrasound imaging.

A participant with a PT classified as clinical and imaging tendinopathy had both sets of features of a PT classified as clinical tendinopathy and imaging tendinopathy.

Participants classified as having a normal PT on left and right sides were included in analyses comparing PT mechanical properties between sides and groups. The PT mechanical properties for participants classified as having either clinical tendinopathy, imaging tendinopathy or clinical and imaging tendinopathy are presented as a case series.

<u>6.3.6.ii Patellar Tendon Mechanical Properties</u>

The derived variables were PT force, CSA, elongation, stiffness, strain, stress and elastic modulus. The method for data processing and calculation of PT force, CSA, elongation and stress were described in chapter 5 (section 5.3).

6.3.6.ii - Normalised Patellar Tendon Force

Patellar tendon force was calculated from measured knee moment (Nm) divided by the measured PT moment arm length (chapter 5, section 5.3). The force transmitted through the PT is proportional to the force generated by the knee extensors. As muscle force is proportional to the physiological cross-sectional area of the muscle, the value of muscle force was normalised to mass raised to the power 2/3 as a size independent measure of muscle strength (Jaric, 2002). This non-linear allometric scaling allows for comparisons between participants of different sizes.

The normalised PT force was calculated to allow comparison between groups of different size and weight. The following equation was used to normalise the PT force values (Jaric, 2002).

$$S_n = S/m^b$$

Where S_n is normalised PT force (N), S is recorded PT force (N) and m is body mass (kg) and b is 2/3.

6.3.6.ii - Stiffness and Elastic Modulus

The force-length and stress-strain curve for each leg for each participant was determined by recording the elongation and strain at each 10% interval of MVIC respectively. The gradient of each 10% MVIC increment (20-30%, 30-40%, 40-50%, 50-60%, 60-70% and 70-80%) for force-length and stress-strain curves were determined and represented the stiffness and elastic modulus respectively. The term ‘stiffness’ in this chapter (6) and the following chapter (7) refers to the slope of the force-length curve of the PT at a 10% increment (20-30%, 30-40%, 40-50%, 50-60%, 60-70% and 70-80%) of the MVIC of the knee extensor muscle group.

6.3.7 Data Analyses

6.3.7.i Clinical Profile

Scores of the VAS and VISA questionnaires were reported for each sporting group with normal bilateral patellar tendons for each side and for each participant with patellar tendinopathy (classified as either clinical tendinopathy, imaging tendinopathy or clinical and imaging tendinopathy) on either one or both sides. For athlete groups with normal patellar tendons bilaterally, the number of participants who reported either the left or right side to be dominant during each function was reported. The dominant side (left or right) for each participant with patellar tendinopathy was reported for each function.

6.3.7.ii Patellar Tendon Mechanical Properties

Group by Side Comparisons

A 2-way ANOVA compared differences between sides (left and right) and groups (men's volleyball players, men's basketball players, women's basketball players and swimmers) for normalised force, proximal CSA, strain and stress. A 2-way ANOVA was used to compare PT stiffness and elastic modulus of the PT at each 10% MVIC interval. Statistical significance was accepted at the 95% confidence limit ($p < 0.05$).

If no significant interaction was detected then the main effect was identified. Pairwise comparisons were performed to identify differences between sporting groups. If there was a main effect for side, multiple paired t-tests were performed post hoc, to identify inter-side differences within each sporting group. Statistical significance was accepted at the 99% confidence limit ($p < 0.01$) after Bonferroni adjustment. All statistical tests were performed using PASW Statistics, version 18. The observed power for each post hoc statistical test was calculated using G*Power 3.1.2 (available as freeware on http://www.psycho.uni-dusseldorf.de).

Dominance Related Comparisons

Multiple independant t-tests were performed to compare PT mechanical properties (elongation, normalised force, stiffness (at 10% intervals from 20-80% MVIC)), strain, stress and elastic modulus (at 10% intervals from 20-80% MVIC) between groups (i.e.; left dominant against right dominant) for each side (left and right) within each sport with significance accepted at $p < 0.05$ (PASW Statistics, version 18). This analysis was only performed on groups where there were at least five samples per group. The observed power for each post hoc statistical test was calculated (G*Power 3.1.2, available as freeware on http://www.psycho.uni-dusseldorf.de).

Patellar Tendinopathy Related Comparisons

Comparisons on PT mechanical properties were made between the patellar tendons (left or right) with tendinopathy to the mean of left and right PT mechanical properties (mean and 95%CI) of the respective group of athletes (men's volleyball players and men's basketball players) who had normal patellar tendons bilaterally. This was a descriptive analysis as inferential statistics could not be used to the small sample of tendons with tendinopathy.

Patellar Tendon Cross-sectional Area

Inter-side comparison of proximal, mid and distal PT CSA was performed using paired t-tests for each sporting group with significance accepted at $p < 0.01$ (PASW Statistics, version 18). A more conservative alpha level was used as multiple comparisons were made. If there was no significant inter-side difference, the mean value between left and right sides for proximal, mid and distal PT CSA was calculated (Microsoft Excel, 2003). A 1-way repeated ANOVA with a factor of site (3 levels for proximal, mid and distal) was performed for comparison across three sites for each sporting group and statistical significance was accepted at $p < 0.05$ (PASW Statistics, version 18).

6.4 Results

6.4.1 Clinical Examination

Fifty one of the 60 participants tested had normal patellar tendons bilaterally. VISA scores were high and VAS scores were low for participants classified as having normal bilateral patellar tendons compared to the 9 participants with patellar tendinopathy on either side (Table 6.4.1).

The right side was the primary functional dominant limb for participants with normal bilateral tendons (Table 6.4.2). Scores for the VISA questionnaire and visual analogue scales were pooled between the 4 male and 2 female swimmers as they all had normal left and right patellar tendons.

Table 6.4.1 Clinical profile of athlete groups with normal patellar tendons bilaterally and participants with patellar tendinopathy.

Sport		Classification		VISA		VAS (1)		VAS (2)		Dominant Side During Function				
		L	R	L	R	L	R	L	R	Write	Shoot/ Spike	Kick	Jump	Land
♂ Volleyball$_{(n=18)}$		N	N	90.6 (11.6)	84.7 (16.9)	17.6 (22.3)	13.9 (22.9)	6.8 (13.0)	17.8 (22.3)					
1	♂ VB	N	I	77	72	5	15	5	4	L	L	L	R	R
2	♂VB	C & I	QT	42	53	71	17	25	27	R	R	L	L	L
3	♂ VB	C	N	57	63	67	19	39	17	R	R	R	L	L
4	♂ VB	C & I	C & I	70	70	81	68	78	37	R	R	L	L & R	L
5	♂ VB	C	C & I	86	86	23	26	0	0	L	L	R	R	R
♂ Basketball$_{(n=13)}$		N	N	82.6 (15.6)	90.2 (10.9)	14.7 (17.1)	6.1 (11.7)	13.1 (21.7)	3.4 (4.8)					
6	♂ BB	C	N	70	92	76	32	81	13	R	R	R	L	R
7	♂ BB	C & I	N	74	92	45	0	15	0	R	R	R	L	L
8	♂ BB	C	C & I	66	63	46	55	8	13	R	R	R	L	L
9	♂ BB	C & I	N	78	98	40	7	23	4	R	R	R	L	L
♀ Basketball$_{(n=14)}$		N	N	89 (15.5)	87.5 (13.6)	12.1 (21.0)	11.0 (17.6)	12.7 (22.1)	10.3 (19.0)					
Swimming$_{(n=6)}$		N	N	90.7 (8.6)	92.3 (7.1)	18.5 (30.9)	12.2 (9.4)	1.3 (3.3)	0.0 (0.0)					

VB – Volleyball; BB – Basketball; VISA – Victorian Institute of Sport Assessment; VAS – Visual Analogue Scale; (1) – VAS during last one week of activity; (2) VAS during single leg decline squat; L - Left; R – Right; QT – Quadriceps Tendinosis

Table 6.4.2 Dominance related details of athlete groups with normal patellar tendons bilaterally.

Group	Writing hand		Shoot/Spike hand		Kicking leg		Jumping leg		Landing Leg	
	Left	Right	Left	Right	Left	Right	Left	Right	Left	Right
Men's Volleyball $_{(n=18)}$	1	17	1	17	8	10	13	5	13	5
Men's Basketball $_{(n=13)}$	1	12	1	12	1	12	11	2	11	2
Women's Basketball $_{(n=14)}$	1	13	0	14	0	14	10	4	8	6
Swimming $_{(n=6)}$	0	6	N/A	N/A	0	6	0	6	2	4

6.4.2 Patellar Tendon Mechanical Properties

Patellar tendon mechanical properties for the 51 participants classified as having normal patellar tendons bilaterally (Table 6.4.3) were similar to the PT mechanical properties for the 9 participants classified as having patellar tendinopathy (Table 6.4.4). Patellar tendon mechanical properties for male and female swimmers were pooled due to the small sample size.

Table 6.4.3 Patellar tendon mechanical properties (mean (SD)) for each sporting group with normal patellar tendons bilaterally.

Group	Normalised Force ($N.kg^{-2/3}$)		Cross-Sectional Area (mm^2)		Stiffness ($N.mm^{-1}$)		Strain (%)		Stress (MPa) ($N.mm^{-2}$)		Modulus (GPa)	
	Left	Right	Left	Right	Left	Right	Left	Right	Left	Right	Left	Right
♂ VB	185.43	199.11	171.84	174.27	2834.77	3215.50	12.07	12.10	22.37	24.08	0.84	1.11
(n=18)	(61.52)	(79.00)	(37.46)	(35.86)	(1232.38)	(1512.30)	(3.63)	(4.00)	(7.63)	(10.95)	(0.45)	(0.93)
♂ BB	154.72	162.12	190.91	197.35	2446.09	2386.11	8.70	9.69	16.20	16.78	0.87	0.67
(n=13)	(57.29)	(68.50)	(33.36)	(45.68)	(1212.65)	(1533.96)	(2.67)	(3.70)	(5.86)	(7.59)	(0.67)	(0.46)
♀ BB	162.55	175.48	200.55	199.76	3669.09	2248.83	11.16	10.63	15.48	16.57	0.79	0.61
(n=14)	(32.37)	(38.26)	(22.54)	(45.92)	(2929.96)	(1354.58)	(2.70)	(2.49)	(4.16)	(4.09)	(0.79)	(0.40)
SW	190.59	218.55	160.08	156.50	1900.00	2537.78	10.87	10.76	21.23	25.27	0.57	0.81
(n=6)	(38.48)	(29.29)	(25.87)	(29.67)	(1098.40)	(1326.00)	(2.72)	(2.06)	(4.65)	(5.58)	(0.30)	(0.43)

VB – Volleyball; BB – Basketball; SW - Swimming

Table 6.4.4 Patellar tendon mechanical properties (mean (SD)) for the left and right sides of men's volleyball and men's basketball groups with normal patellar tendons bilaterally and each participant classified as having patellar tendinopathy.

		Side	Classification	Normalised Force ($N.kg^{-2/3}$)	Proximal Cross-Sectional Area (mm^2)	Stiffness† ($N.mm^{-1}$)	Strain (%)	Stress (MPa)	Elastic Modulus† (GPa)
Volleyball$_{(n = 18)}$		L	N	185.43 (61.52)	171.84 (37.46)	2834.77 (1232.38)	12.07 (3.63)	22.37 (7.63)	0.84 (0.45)
		R	N	199.11 (79.00)	174.27 (35.86)	3215.50 (1512.30)	12.10 (4.00)	24.08 (10.95)	1.11 (0.93)
1	VB	L	N	227.75	136.7	2021.5	12.83	36.45	0.70
		R	I	202.41	226.7	1209.99	15.71	19.53	0.25
2	VB	L	C & I	200.47	209.9	870.12	16.09	19.76	0.22
		R	QT	194.11	182.4	2574.62	10.81	22.017	0.78
3	VB	L	C	114.48	221.2	2260.55	6.42	9.83	0.49
		R	N	110.52	186.7	5644.00	4.87	11.25	1.76
4	VB	L	C & I	96.41	199.1	2595.32	6.57	10.52	0.73
		R	C & I	40.14 (<2SD)	236.6	2769.11	5.66	3.69 (<2SD)	0.59
5	VB	L	C	129.01	196.0	1577.07	6.66	12.91	0.47
		R	C & I	251.54	209.2	3639.88	5.14	23.59	0.92
Basketball$_{(n = 13)}$		L	N	154.72 (57.29)	190.91 (33.36)	2446.09 (1212.65)	8.70 (2.67)	16.20 (5.86)	0.87 (0.67)
		R	N	162.12 (68.50)	197.35 (45.68)	2386.11 (1533.96)	9.69 (3.70)	16.78 (7.59)	0.67 (0.46)
6	BB	L	C	139.75	253.8	2035.44	15.65	11.23	0.30
		R	N	162.70	256.4	2015.21	10.05	12.94	0.56
7	BB	L	C & I	141.18	166.5	1237.97	13.11	18.92	0.35
		R	N	170.74	160.9	2166.20	9.74	23.68	0.70
8	BB	L	C	122.85	182.0	1224.52	10.55	12.72	0.29
		R	C & I	158.91	156.4	506.82	15.44	19.14	0.14
9	BB	L	C & I	128.71	274.6 (>2SD)	1954.13	10.71	9.98	0.38
		R	N	220.20	277.1 (>2SD)	2509.51	11.51	16.92	0.59

VB – Volleyball; BB – Basketball; L – Left; R - Right; N – Normal Patellar Tendon; I – Imaging Patellar Tendinopathy; C – Clinical Tendinopathy; † - calculated from 70 to 80% MVIC; <2SD – Less than 2 SD from the mean of the comparative group of athletes with normal patellar tendons bilaterally; >2SD – Greater than 2SD from the mean of the comparative group of athletes with normal patellar tendons bilaterally

6.4.2.i - Normalised Patellar Tendon Force

There was a significant main effect for side (p = 0.011) and no main effect for group (p = 0.204) (Appendix N: Figure N.1). Thus, the right side had greater normalised PT force than the left in all athlete groups, however, post hoc multiple paired t-tests detected a significantly greater normalised PT force on the right compared to the left side in the swimming group only (p = 0.046) (Figure 6.4.1) (Appendix N: Table N.1).

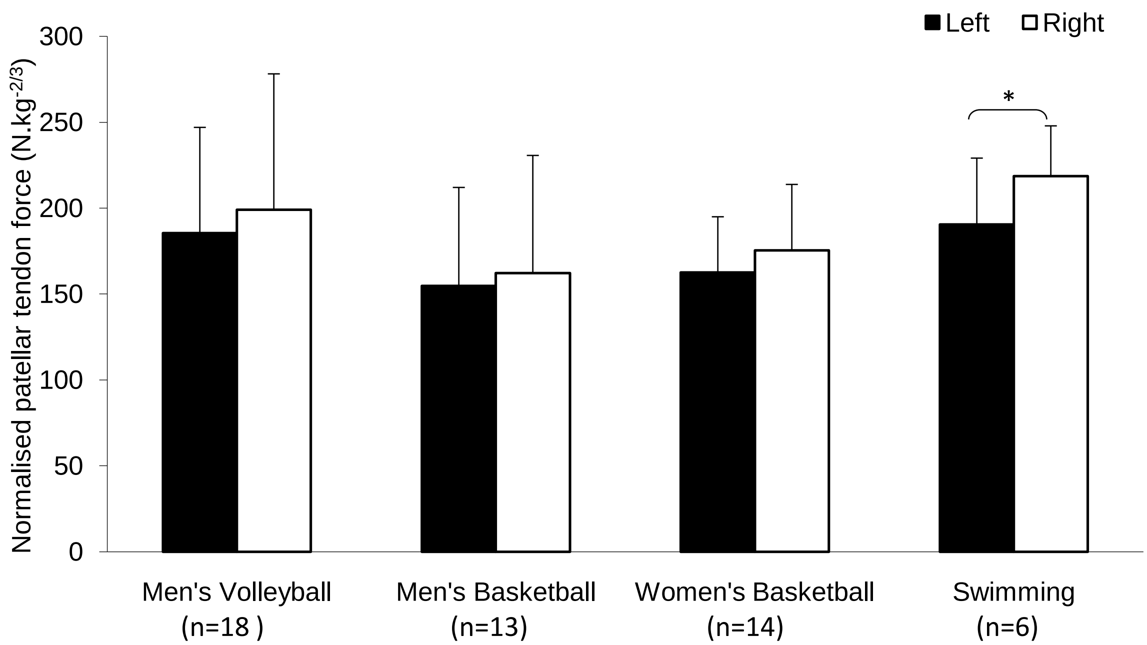


Figure 6.4.1 Normalised patellar tendon force (N.kg$^{-2/3}$) (mean SD) is significantly greater on the right than left side in the swimming group (*p < 0.05).

6.4.2.ii Proximal Patellar Tendon Cross-Sectional Area

There was no significant 2-way interaction (p = 0.91) or main effect for side (p = 0.82) or group (p = 0.91) (Appendix N: Figure N.2).

6.4.2.iii Patellar Tendon Strain

There was no significant 2-way interaction (p = 0.36) or main effect for side (p = 0.78) or group (p = 0.09) (Appendix N: Figure N.3).

6.4.2.iv Patellar Tendon Stiffness

There was no significant interaction ($p \geq 0.05$) or main effect for side ($p \geq 0.05$) or group ($p \geq 0.05$) when comparing stiffness at any 10% interval from 20 to 80% MVIC (Appendix N: Table N.2). Patellar tendon stiffness for each group at each 10% MVIC interval and an example of a force-length curve for the left and right side of a single participant are shown in Figures 6.4.2 and 6.4.3 respectively.

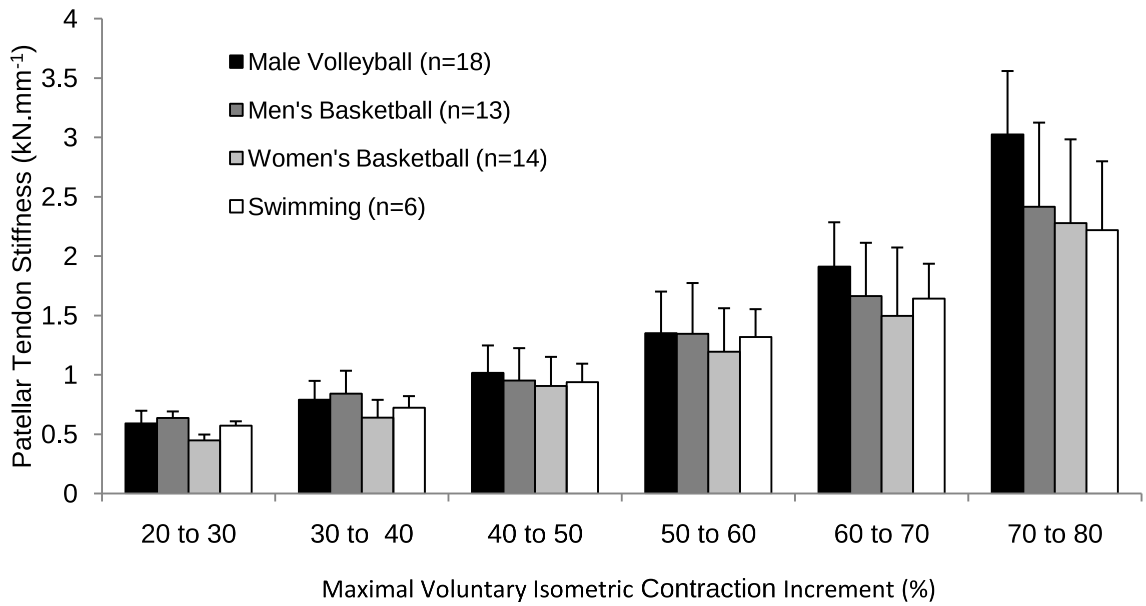


Figure 6.4.2 Patellar tendon stiffness (kN.mm^{-1}) (mean SD) is not significantly different between groups at 10% intervals from 20% to 80% maximal voluntary isometric contraction effort.

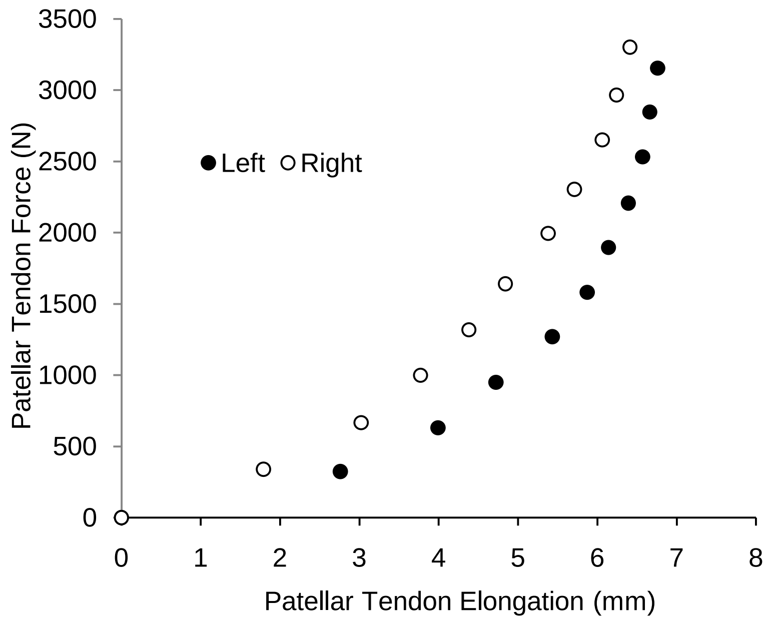


Figure 6.4.3 Force-Length curve for the left and right side of a single participant (men's volleyball player) from 0% to 100% maximal voluntary isometric contraction effort.

6.4.2.v Patellar Tendon Stress

There was no significant interaction between side and group ($p = 0.58$) with significant main effects for side ($p = 0.02$) and group ($p = 0.007$) when comparing PT stress (Appendix N: Figure N.4). Pairwise comparisons detected significantly greater PT stress in the men's volleyball group compared to the men's basketball ($p = 0.049$) and women's basketball ($p = 0.025$) groups (figure 6.4.4) (Appendix N: Table N.3).

The right side had greater PT stress than the left for all athlete groups, however, post hoc multiple paired t-tests detected no significant inter-side difference in normalised PT force in any sporting group. In the swimming group the difference was greatest with the right PT having greater PT stress than the left ($p = 0.024$) (Figure 6.4.4) (Appendix N: Table N.4).

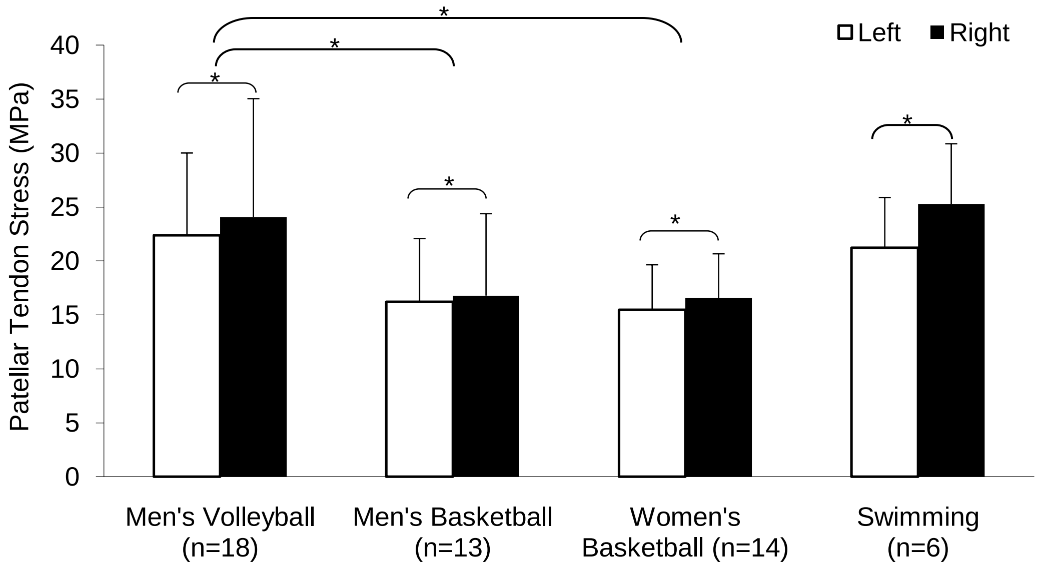


Figure 6.4.4 Patellar tendon (PT) stress (MPa) (mean SD) is significantly greater in the male volleyball group compared to the male and female basketball groups (*). There is a significantly greater PT stress on the right compared to left side in all athlete groups (*p < 0.05).

6.4.2.vi Patellar Tendon Elastic Modulus

There was no significant interaction between side or group ($p \geq 0.05$) or main effects for either side ($p \geq 0.05$) or group ($p \geq 0.05$) when comparing PT elastic modulus at any 10% interval from 20 to 80% MVIC (Appendix N: Table N.5).

6.4.2.vii Normal Patellar Tendons - Dominance Related Comparisons

Men's volleyball players stratified by the preferred landing leg showed that the left PT of participants that preferred to land on their left leg had significantly lower stiffness than the left PT of those that preferred to land on their right leg, at the 70 to 80% MVIC interval ($p = 0.036$) with mean 2466N.mm^{-1} (SD) (997) and 3793N.mm^{-1} (1374) for each group respectively (Figure 6.4.5) (Appendix N: Table N.6). There was no difference in PT stiffness of the right side for groups stratified by preferred landing leg ($p \geq 0.05$) (Figure 6.4.5) (Appendix N: Table N.6).

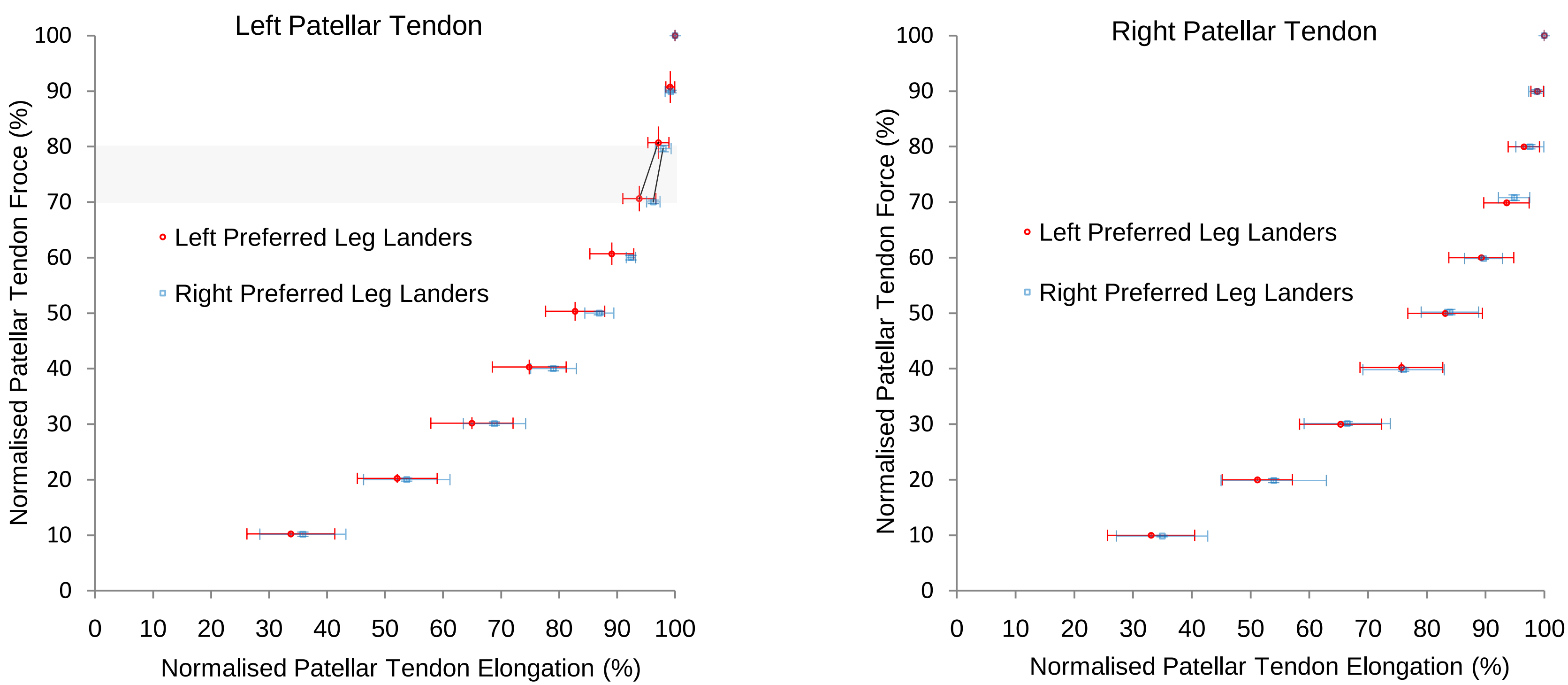


Figure 6.4.5 Normalised force-length (mean SD) curves for men's volleyball players. Patellar tendon (PT) stiffness at the increment from 70 to 80% maximal voluntary isometric contraction effort (slope shown in solid black line) was significantly lower for the left PT of those that preferred to land on their left leg than the left PT of those that preferred to land on their right leg (figure on left) for PT force from 70 to 80% of maximal voluntary isometric contraction (shaded grey area). Stiffness of the right PT for players that preferred landing on either their left or right was similar (right figure).

There was no significant difference for elastic modulus between groups stratified by the preferred landing leg in men's volleyball players on the left ($p \geq 0.05$) or right ($p \geq 0.05$) (Appendix N: Table N.7).

There were no other significant differences in any sporting group when stratified by the preferred landing leg, jumping leg, shooting/spiking hand or writing hand.

6.4.2.viii Patellar Tendinopathy Related Comparisons

Nine participants were classified as having patellar tendinopathy (Table 6.4.1). All participants classified with patellar tendinopathy were men's volleyball or men's basketball players. Eight of the 9 participants had patellar tendinopathy on the left side and three of these 8 participants had patellar tendinopathy on the right side also.

The normalised PT force, strain, stiffness, proximal CSA, stress and elastic modulus values for left and right tendons with tendinopathy in the men's volleyball and basketball participants were within the 95%CI range for the mean of left and right tendons of men's volleyball and basketball groups that had normal patellar tendons bilaterally (Table 6.4.4) (Appendix N: Figure N.5a-f).

6.4.2.ix Patellar Tendon Cross-Sectional Area - Normal Patellar Tendons

There was no inter-side difference for PT CSA at either the proximal, mid or distal sites of measurement for any sporting group ($p \geq 0.1$) (Figure 6.4.6) (Appendix N: Table N.8). Only swimmers had significantly greater PT CSA at the proximal site compared to the mid PT measurement site ($p = 0.01$) (Figure 6.4.6) (Appendix N: Table N.9 and N.10).

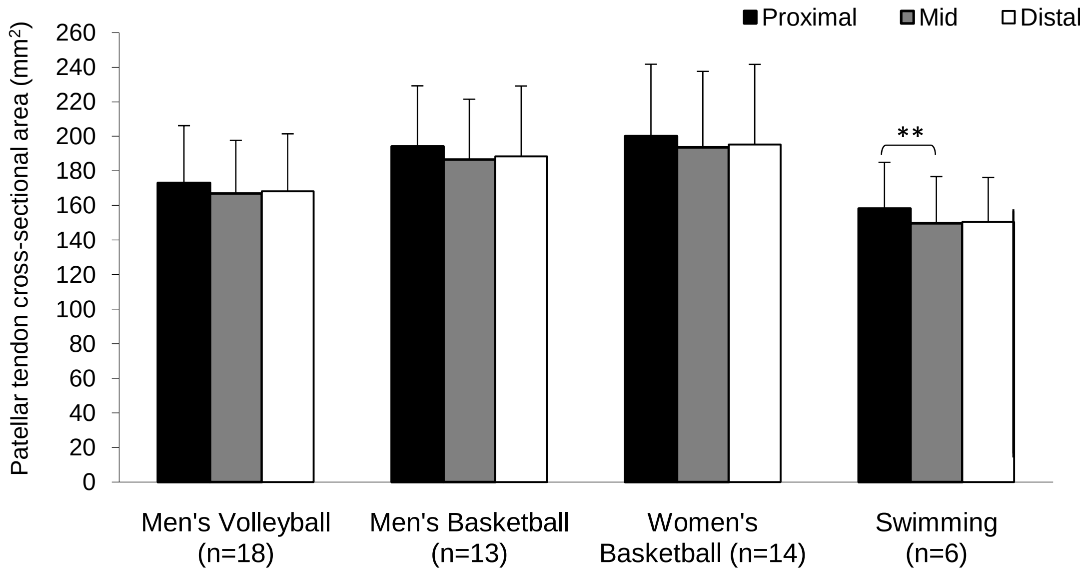


Figure 6.4.6 Patellar tendon (PT) cross-sectional area (mm^2) (mean SD) of normal patellar tendons was significantly greater at the proximal compared to the mid PT measurement site for the swimming group (p < 0.01).**

6.5 Discussion

6.5.1 Patellar Tendon Mechanical Properties

<u>6.5.1 Normal Patellar Tendons</u>

This study demonstrates that the PT mechanical behaviour under load, across a range of elite level jumping and non-jumping athletes with normal tendons is very similar. The single difference was that the left PT of men's volleyball players who prefer to land on their right leg had greater PT stiffness (70 to 80% MVIC) than the left PT of men's volleyball players who prefer to land on their left leg. This difference in PT stiffness was only evident at near maximal effort of knee extensor contraction from 70 to 80% MVIC effort. The significant difference in PT stiffness was due to the left PT elongation, measured at the increment from 70 to 80%MVIC, being lower in the players that preferred to land on their right side compared to players that prefer to land on their left leg, with no difference in normalised PT force.

All activities including sports-specific training, weight training and ADL will influence the patellar tendon's ability to adapt. The current study demonstrated that that there was an association between leg loading behaviour and PT mechanical behaviour in the men's volleyball group. There was no significant difference in right PT stiffness demonstrated between left and right preferred leg landing groups for men's volleyball players or for either left or right PT stiffness in women's basketball players based on dominance. Comparison of dominance related factors was unable to be made in the men's basketball group as there were unequal sample populations when stratified by the dominant side in functional activities. Thus, the finding in men's volleyball players of a significant difference in left PT stiffness only may be defined as being sport-specific to men's volleyball and task-specific to an effect of laterality for the landing leg.

In the current study, 16 of 17 men's volleyball players with normal patellar tendons bilaterally reported a preference of using the right hand to spike the ball. In volleyball players it was demonstrated that landing from a spike jump for right-handed participants, from differing take-off distances, resulted in a consistent finding that the centre of pressure was greater towards the left foot, with left trunk lean at initial contact (Marquez, et al., 2009). It was also demonstrated that the ankle, knee and hip joints were more flexed on the left than the right and had consistently lower variability for knee joint excursions across trials. Marquez et al. (2009) suggested that this asymmetrical landing behaviour in which the left leg joints were more consistently flexed, may be important in determining absorption of kinetic energy in landing from a spike jump in volleyball (Marquez, et al., 2009). This landing behaviour may manifest itself in a difference in tissue response in a number of ways. The left PT may be 'over-loaded' and may adapt. Conversely, the contralateral side is not being exposed to as great an intensity of forces as the left and may be seen to be relatively 'under-loaded' and may adapt. The process of tendon tissue adaptation is suggested as a possible contributing factor to the pathogenesis of patellar tendinopathy rather than as a process of degeneration or failed healing that are terms used to describe the pathophysiology of tendinopathy (Hamilton & Purdam, 2004).

The current finding of an association between the preferred landing leg and PT stiffness is supported by other studies that demonstrate a relationship between loading behaviour and tissue injury. A previous finding demonstrated increased weight lateralisation between sides and abnormal left and right Q-angles, which were predictive of injury in a group of elite basketball players (Shambaugh, Klein, & Herbert, 1991). Shambaugh et al. (1991)

hypothesised that these factors contribute to a change in loading of the kinetic chain and possibly contributing to injury (Shambaugh, et al., 1991).Similarly, increased postural sway during single-leg stance was predictive of ankle sprain injury in a cohort of 210 high school basketball players (McGuine, Greene, Best, & Leverson, 2000) with no differences based on leg dominance. This supported other studies that determined symmetry between dominant and non-dominant sides, when single-leg landing from a drop jump (Wilkstrom, Tillman, Kline, & Borsa, 2006) and single-leg standing (Hoffman, Schrader, Applegate, & Koceja, 1998). This demonstrates that the choice of left or right foot for balance is task-dependent rather than due to an overriding preference to use the left side, which is preferred for postural stabilisation (Dargent-Paré, Agostini, Mesbah, & Dellatolas, 1992; Gabbard & Iteya, 1996b; Hart & Gabbard, 1997; Whittington & Richards, 1987). It is difficult to establish asymmetry as causative of injury, as tissue change may be adaptive or deleterious. However, the current study suggests that task-specific asymmetry may be associated with a change in tissue mechanical properties. Therefore, the association between laterality and tissue adaptation may be one that is dependent on task-specific laterality, rather than global asymmetry in static and dynamic postures.

This study also demonstrated that the *in vivo* mechanical behaviour of a PT with either clinical or imaging patellar tendinopathy is similar to normal patellar tendons that have no pain or sign of tissue disease. The current findings support a recent study that demonstrated that the PT stiffness and modulus were similar in one person with one normal PT and one abnormal PT (Kongsgaard, et al., 2009). Kongsgaard et al. (2009) also demonstrated that PT mechanical properties for patellar tendons with tendinopathy remained the same following a 12 week intervention program that was treated using either cortisone injections, heavy slow exercise program or an eccentric exercise program (Kongsgaard, et al., 2009).

In the current study there may be a number of reasons why PT behaviour under load was influenced only in the men's volleyball group and only observed on the left side aside from the possibility it may be a type 1 error. Research findings identify the ipsilateral leg to the dominant hand as the leg with which individuals will perform a dynamic task (Beling, et al., 1998) such that the static or support task is performed by the contralateral leg to the dominant hand (Gabbard, et al., 1996a). This cross-lateral dominance pattern demonstrates a lack of independence between hand and foot laterality during function (Olex-Zarchyta & Raczek, 2008) and is a more consistent finding in right-handed compared to left handed individuals (Beling, et al., 1998). In the current investigation, 17 of the 18 men's volleyball players with normal patellar tendons bilaterally reported a preference for using their right hand as the spiking hand in volleyball. Thus, it may be suggested that the left leg will likely be the leg used for initial support in landing. However, if the preferred landing leg is not the left, then the player may be forced to use the left leg when it is the non-preferred landing leg. This may be more prevalent in volleyball where jumping and landing strategy may be more specific to right-hand dominance. Therefore, landing on the left leg may follow this preferred pattern of spiking with the right hand and the tissue response may differ between sides that may behave differently when sidedness of foot and hand function is not constrained. Future studies may aim to determine the preferred landing leg by volleyball players during training and competition, as well as measuring the PT mechanical properties, to be able to associate motor behaviour and intrinsic tissue properties.

The finding of greater stiffness of the left PT in the group who preferred to land on the right may be a result of tissue adaptation relative to the side that may be exposed to greater intensity of loading. Assessing inter-side asymmetry from a 60cm drop landing, which would be expected to be a equally distributed in a double-leg landing, demonstrated the knee joint had more joint moment asymmetries than the ankle and hip joints (Schot, et al., 1994). Schot et al. (1994) reported that the linear kinetic variables related to force generated at landing were relatively stable compared to individual joint moment variations (Schot, et al., 1994). Hence, there may be an inherent asymmetry at the level of the joint and its ability to generate force that is dependent on MTU function. If conditions of landing are such that landing is dominated by a preferred landing leg, such as in landing from a spike or block jump in volleyball (Marquez, et al., 2009; Tillman, et al., 2004), landing behaviour may predicate asymmetrical tissue loading.

The cross-sectional design of this study allows an association to be made between landing leg and PT stiffness; however it does not identify the causative factor of the difference in PT stiffness. Furthermore, in this study the magnitude of the difference in PT stiffness may bot be clinically significant, however, does require future investigations to determine whether tendon stiffness is influenced by task-specific laterality. Tissue adaptation will be influenced by all activity including ADL and sports-related training. It may be that landing in volleyball forces the player to land on their left leg regardless of preference, as they prefer to use the right hand in spiking. During ADL it is likely that the magnitude of tensile force is lower than compared to landing in volleyball. However, those that prefer to land on their left leg may have an invariant bias to load the left leg during training and during ADL, while those that prefer to land on their right leg may have a bias to load the left leg only during training and not during ADL. The adaptive response of the PT to the volume and intensity of tissue loading is linked; however, the loading characteristics of the lower limb that lead to changes at the local tissue level of the PT remain complex.

In the current study, no significant inter-side difference was detected for measures of PT CSA, strain, stiffness and elastic modulus in this study on elite jumping and non-jumping athletes consistent with a previous study on PT mechanical properties (Kongsgaard, et al., 2007). Patellar tendon stress was greatest on the right compared to the left side with this inter-side difference the largest in the swimming group. This is likely a result of the swimming group having the greatest inter-side difference in PT force while the proximal PT CSA was relatively symmetrical between sides. Therefore, knee extensor strength asymmetry may explain the inter-side difference in PT stress in swimmers, however, the small sample size (n=6) limits interpretation.

The current findings of relative knee extensor strength symmetry in jumping athletes compared to the non-jumping athletes supports previous studies that also detected a trend for the dominant leg (preferred kicking leg) to have greater knee extensor strength in a non-athletic population, while jumpers and runners were relatively symmetrical when tested isokinetically at slow (60º/second) and fast (240º/second) speeds (Siqueira, Pelegrini, Fontana, & Greve, 2002). Swimmers do not expose their knee extensors to the same volume or intensity of activity as jumping athletes and hence may maintain an asymmetry that is inherent amongst non-athletes. In a healthy, untrained and uninjured population, strength asymmetry may be a ‘normal’ finding or a product of task specific-dominance. Although not the primary purpose of this investigation, it may be asked whether elite swimming lends itself to knee extensor strength asymmetry due to training effect or that a finding of a difference of the right knee extensors being 16.3% greater than the left is a ‘normal’ finding. By contrast, jumping athletes in this study had greater PT force on the right than left ranging from between 5% to 10% and although the main effect was significant, it represents an inter-side difference within the accepted range for inter-side difference (<10%) in athletic and non-athletic populations (Couppé, et al., 2008; Rahnama, Lees, & Bambaecichi, 2005; Schiltz, Lehance, Maquet, Bury, Crieland, & Croisier, 2009).

In the current study, jumping athletes who do participate in activities that may load the legs asymmetrically (Schot, et al., 1994; Tillman, et al., 2004) were found to be symmetrical in measures of PT CSA, stiffness, strain and elastic modulus. Basketball and volleyball may be considered to include motor tasks that use each leg equally such as in running, double-leg jumping and landing as well as those that favour one side such as in a lay-up, serving and spiking.

There may be two postulates when comparing elite jumping athletes to athletes that do not participate in jumping activities to explain the relative symmetry of PT force, CSA and mechanical properties: First, there is maintenance of symmetry, second, there is adaptive change that leads to symmetry. There was a relatively small inter-side knee extensor strength difference detected in this investigation compared to a previous investigation of elite fencers and badminton players (Couppé, et al., 2008). Couppé et al. (2008) found 7 of 22 participants in these sports to have a ≥15% inter-side quadriceps femoris strength difference and suggested that the findings of greater PT stiffness and CSA in the stronger, lead extremity comply with the notion that the PT may adapt in athletes that habitually load their knee extensors in an eccentric manner.

In the current study there may not be a great enough knee extensor strength asymmetry (~13%) in the jumping athletes to lead to asymmetry in mechanical properties. Couppé et al. (2008) excluded 15 of 22 recruits from testing of PT mechanical properties, as their knee extensor strength asymmetry was <15%. However, these 15 elite level athletes were selected from the same population as the 7 elite level athletes that had a knee extensor strength asymmetry of >15% and found to have greater PT stiffness of the stronger extremity (Couppé, et al., 2008). It is an assumption that loading behaviour of the PT and therefore tissue adaptation is only responsive to ‘over-loading’ relative to the contralateral side. In the current study, the left PT that is potentially the side to be ‘over-loaded’ was stiffer in those who did not prefer to land on the left side. Thus, it is possible that the significant difference in PT mechanical properties detected by Couppé et al. (2007) on elite level fencers and badminton players may be a result of differences in loading behaviour based on the lead and non-lead leg, rather than knee extensor strength asymmetry and that there may have been differences in PT mechanical properties in the sample population with a knee extensor strength asymmetry of <15%. Therefore the asymmetry in PT mechanical properties may not be due to a difference in the strength of the muscle acting about the joint but may be due to a difference in the loading of the joints. The question remains as to whether the trend for knee extensor strength asymmetry in land-based and aquatic sport at the elite level is inherent or a training effect. The relationship between function, limb loading and tissue adaptation are complex and may not be simply observed as a product of tissue over-load.

While there was no significant difference between sporting groups for normalised PT force or CSA, PT stress (force/CSA) was significantly lower in the men’s basketball and women’s basketball groups compared to the men’s volleyball group by 28.9% and 31.0% respectively. This is likely a product of the men’s and women’s basketball groups both having lower PT force measures (17.6% and 12.1% respectively) and greater PT CSA measures (12.2% and 15.7% respectively) compared to men’s volleyball players. Importantly, the asymmetry in PT stress detected during the isometric contraction may be different than that experienced during sport, possibly limiting the ecological validity.

The values for proximal PT CSA in this study (158 to 194mm^2 in males and 200mm^2 in females) were generally higher than those reported in previous studies (73 to 163mm^2 for males and 63 to 91mm^2 for females) (Couppé, et al., 2008; P. Hansen, et al., 2006; O'Brien, et al., 2010; Westh, et al., 2008). This is due to a number of reasons. This study examined elite level athletes who trained full-time in each sport rather than recreational athletes. As increased PT CSA is shown to occur in response to only 12 weeks of strength training (Kongsgaard, et al., 2007) it is likely that athletes training in full-time capacity have greater PT CSA than recreational athletes or untrained recruits. The finding of greater PT CSA was consistent in both jumping and non-jumping athletes and in both genders.

Methods for the measurement of PT CSA have included the use of MRI (Couppé, et al., 2008; Kongsgaard, et al., 2007; Kubo, et al., 2006; Westh, et al., 2008) and ultrasound (Burgess, et al., 2009a; P. Hansen, et al., 2006; Maganaris, et al., 2004; O'Brien, et al., 2010; Onambélé, et al., 2007a; N. D. Reeves, et al., 2003a) with a strong correlation between both techniques (N. R. Reeves, et al., 2009). Care was taken to ensure that the borders of the PT were identified by recording a series of real-time ultrasound images that allowed the transducer to be moved along the transverse plane of the PT. Thus, when measuring the PT CSA from a single image, the image series was scrolled back and forth from the single frame used for measuring the CSA. This method allowed clear identification of PT borders that may be unclear on a single image only.

There was a trend for athletes participating in jumping sports to have greater PT CSA compared to swimmers. A study of the Achilles tendon revealed runners and volleyball players to have a larger normalised Achilles CSA than athletes of non-weight bearing sports (kayakers) (Kongsgaard, Aagaard, Kjaer, & Magnusson, 2005) and for male runners to have greater Achilles tendon CSA than non-runners (Rosager, Aagaard, Dyhre-Poulsen, Neergaard, Kjaer, & Magnusson, 2002). In the current study there was no statistically significant difference between the three jumping groups (men’s volleyball, men’s basketball and women’s basketball) with these groups having a relatively equal number of participants in each group. The swimming group only had a total of six participants and may not be representative of a larger population of swimmers.

The range of values for PT force in this study from mean of 2994N (SD) (58) to 3811N (149) were lower than previously reported (~2000N to ~11000N) (Burgess, et al., 2009b; S. J. Pearson, et al., 2005; S. P. Pearson, et al., 2007). This study did not estimate the antagonist (hamstring) muscle contribution to the recorded knee extensor moment (see section 5.5.1) Thus, the values for PT force are likely to be lower than studies that have utilised a sum of extensor torque and correction for flexor torque to calculate PT force (Bojsen-Møller, et al., 2003; Burgess, et al., 2009c; S. J. Pearson, et al., 2005; N. D. Reeves, Narici, & Maganaris, 2004; N. R. Reeves, et al., 2009; Westh, et al., 2008). The magnitude of force contribution of antagonistic hamstring activity may be indeterminate from EMG measurement of activation amplitude when antagonist moment is unknown and was not measured in this study due to these methodological factors (Kellis, 1998). Furthermore, increase in hamstring EMG activity is shown to not be proportional to the increase in extension moment measured quasi-isometrically from 5º to 50º knee flexion (Kingma, et al., 2004) and co-activation is shown to vary across the medial and lateral parts of the hamstring (Aagaard, Simonsen, Andersen, Magnusson, Bojsen-Moller, & Dyhre-Poulsen, 2000). Therefore, correction of antagonist contribution to knee extensor torque, based on EMG activation values of one or more hamstring muscles, may not provide an accurate estimate of correction for knee extension moment. Furthermore, shorter (3s) duration of contraction were shown to generate greater measures of MVIC than longer duration of contraction (10s) (S. P. Pearson, et al., 2007). Thus, the use of slow contractions (10s) in this study may also contribute to the findings of lower PT force.

The measure of PT moment arm length in this study (chapter 5) was participant-specific and measured using a direct geometric method approximating the position of testing with the knee in 90º of flexion. Values in this study ranged from a mean of 45mm (SD) (4) to 55mm (3) and were generally at the upper limit of values in other studies that used an average of values reported from previous studies (Burgess, et al., 2009b; Burgess, et al., 2009c; Onambélé, et al., 2007a; Onambélé, et al., 2007b), estimates based on a cadaveric study (Couppé, Hansen, Kongsgaard, Kovanen, Suetta, Aagaard, Kjaer, & Magnusson, 2009; Couppé, et al., 2008; P. Hansen, et al., 2006; Kongsgaard, et al., 2009; Kongsgaard, et al., 2007; Kubo, et al., 2006; Visser, et al., 1990; Westh, et al., 2008) or used MRI (tibio-femoral contact point as the knee axis of rotation) (Maganaris, et al., 2004; O'Brien, et al., 2010; N. D. Reeves, et al., 2003a; N. D. Reeves, et al., 2003b; N. R. Reeves, et al., 2009). Therefore, greater values of PT moment arm will yield a lower value for PT force than previously reported. Furthermore the PT moment arm length was measured to the anterior margin of the PT and the PT length was measured at the posterior margin of the PT. This may be a source of error in the calculation of PT force in the current study. Other than methodological differences in the estimate of PT moment arm length to previous studies, the greater values in the current study may be explained by the greater stature of participants (Table 6.3.1) (Maganaris, et al., 2005) that included elite level jumping athletes who have greater height than the general population. Although the factors described contributed to the lower values for PT force in this study compared to previous studies, the currently used method was common for all participants being tested. This specifically relates to the gradient of the force-length and stress-strain curves being calculated with respect to the percentage effort from 0 to 100% of MVIC at 10% intervals.

Strain was not significantly different between sides or groups in the current investigation (Table 6.4.3) and minimum and maximum values of 3.4% and 17.7% respectively for normal patellar tendons. For participants with patellar tendinopathy the minimum and maximum PT strain values were 4.9% and 16.1% indicating that strain values were similar between normal and abnormal patellar tendons. Considering PT strain was relatively symmetrical between sides for all sporting groups, these values are within the range of strain values reported in previous findings (Table 6.5.1). Strain values investigated *in vivo* range from ~6.0% to 16.5% in young men (Carroll, Dickinson, Haus, Lee, Aagaard, Magnusson, & Trappe, 2008; P. Hansen, et al., 2006; Kongsgaard, et al., 2006; Kongsgaard, et al., 2007; Kubo, et al., 2005a; Maganaris, Reeves, Rittweger, Sargeant, Jones, Gerrits, & De Haan, 2006; Onambélé, et al., 2007a; Onambélé, et al., 2007b; S. P. Pearson, et al., 2007; Sheehan, et al., 2000; Westh, et al., 2008), 4.8% to 15.3% in young women (Carroll, et al., 2008; Kongsgaard, et al., 2006; Onambélé, et al., 2007a; Westh, et al., 2008), 5% to 10.3% in older women (Carroll, et al., 2008; N. D. Reeves, et al., 2003a) and 9.3% in spinal cord injured patients (Maganaris, et al., 2006). The differences in reported strain values are likely due to the varying methods of PT elongation measurement (see chapter 5), participants' age, participants 'activity level and speed of knee extensor muscle contraction.

In the current study the assessment of PT elongation aimed to control for some of these factors. This included using a ramped isometric contraction (10 seconds duration) following a warm-up and familiarisation period that aimed to reduce the effect of viscoelastic creep and control the speed of contraction (S. P. Pearson, et al., 2007). A healthy, elite, level full-time athlete population was recruited. Patellar tendon elongation was measured as a sum of proximal and distal segments of the posterior portion of the PT (chapter 5) rather than the distance between the patella and the shadow of an external marker, or the distance between the patella and tibia, that may not be specific to the attachment sites of the posterior portion of the PT (chapter 5) (Burgess, et al., 2009a; Burgess, et al., 2009b; Kongsgaard, et al., 2006;

Kubo, et al., 2006; Maganaris, et al., 2004; S. J. Pearson, et al., 2005, 2006; N. D. Reeves, et al., 2003a).

Stiffness and elastic modulus ranged from mean 2218Nmm^{-1} (SEM) (190) to 3025Nmm^{-1} (59) and 0.7GPa (0.1) to 2.4GPa (0.1) respectively. These are lower than previously reported values. The previous studies have calculated stiffness and elastic modulus either at the final 90 to 100% MVIC interval (Couppé, et al., 2008; P. Hansen, et al., 2006; Kongsgaard, et al., 2007; Westh, et al., 2008) or at a common PT force chosen from the greatest common force (weakest participant) (Burgess, et al., 2009b; Couppé, et al., 2009; Onambélé, et al., 2007a; S. J. Pearson, et al., 2005). In the current study, stiffness and elastic modulus values calculated from either 80 to 90% or from 90 to 100% MVIC effort ranged from a mean of 2943Nmm^{-1} (SD) (98) to 7914Nmm^{-1} (461) and 0.7GPa (0.1) to 2.4GPa (0.1) respectively. These 'maximal' effort values are similar to the previously reported data on PT mechanical properties.

A significant difference in this study compared to previous studies was that stiffness and elastic modulus were calculated for each participant at a common percentage effort of the MVIC. This is in contrast to previous investigations that have calculated stiffness and elastic modulus at a common force for comparison between sessions (P. Hansen, et al., 2006), subjects (de Boer, et al., 2007; Maganaris, et al., 2006), groups (Carroll, et al., 2008) and sides (Couppé, et al., 2008). By choosing the PT force that was the lowest amongst a group sample, such that all samples may be included, was in effect a sub-maximal effort for all but one participant. This may not reflect the same relative PT loading during function that may be greater than the point at which stiffness is calculated. Therefore, the differences between studies in the method to calculate stiffness and elastic modulus make it problematic to compare results.

Since the force-length relationship is curvilinear, attempts to assess stronger and weaker participants at the same absolute load may underestimate the slope of the force-length and stress-strain curves for participants that are stronger. The gradient of the curve that defines stiffness and modulus may be calculated at mid-range or even the toe-region for stronger individuals relative to a weaker participant and may not be seen as a valid comparison between participants when considering the role of a loaded tendon during function.

There are two reasons for calculating stiffness and elastic modulus at 70 to 80% MVIC effort rather than previously described methods. First, there was no observable PT elongation from 80 to 100% MVIC effort in 37 of 120 legs tested. Hence, stiffness (force/elongation) and elastic modulus (stress/strain) would be a division by zero within this range of PT force. Patellar tendon elongation was within the range of typical error of 0.1mm (95%CI) (0.1 to 0.5mm) for 238 of 240 measurements made from 80 to 90% and 90 to 100%MVIC for all 60 participants included in the study. Thus, it is possible that there was minimal or no PT elongation towards 100% MVIC efforts, determining that the gradient of the slope is undefined for force-length and stress-strain curves. Second, the PT force recorded at 70-80% effort was achieved by all athletes and therefore was more repeatable than the force output at repeated 100% MVIC efforts. This was a critical feature of the methods used in this study as two independent trials were required for all participants for recording proximal and distal segments and the corresponding knee moment (see chapter 5). The fact that the force and elongation data were recorded for trials up to 100% MVIC and then the specific force selected from the lower of the two force recordings (proximal and distal segment) meant that the major source of error attributable to the participant's performance was the speed of force production rather than the ability to perform repeated 100% MVIC efforts. The rate of force production was controlled using a familiarisation period and feedback (visual and verbal) to ensure a graduated force production over 10 seconds for each trial. In a prospective design, as there may be a change in PT force production due to a change in knee extensor strength, stiffness and elastic modulus as a percentage of MVIC effort may be calculated. The change in stiffness and elastic modulus over time may be more reflective of a functional change at the same effort rather than at a fixed force value.

The straightening of the crimp observed in tendons has been described as 'functionally unimportant' at the toe-region of the stress-strain when forces are relatively low (Ker, 2002). Crimp pattern disappears in the early stages of axial strain (<3% strain) in rat tail tendons, pig flexor tendons (K. A. Hansen, Weiss, & Barton, 2002) and rabbit patellar tendons (<67N).

Thus, assessing the gradient of the stress-strain or force-length relationship at efforts greater than 20% allows for a comparison of the gradient of the ‘loading curve’ (Ker, 2002).

Elastic modulus of the PT measured at near-maximal effort (70 to 80% MVIC effort), demonstrated that human PT behaves similarly in athletes that participate in different sports. Elastic modulus (ratio of force/CSA to elongation/resting length) of the PT may be considered a ‘normalised’ value of the ability of the PT to withstand tensile force when the knee extensors contract. This study supports the notion that PT mechanical properties under load for elite jumping and non-jumping athletes are similar to untrained individuals or recreational athletes.

This investigation did not find a difference in PT mechanical properties between men and women basketball players in contrast to a previous study that reported male runners to have greater PT stiffness than females (runners and non-runners) (Couppé, et al., 2009). In recreationally active adults, males were reported to have greater PT CSA, stiffness and elastic modulus (calculated at maximal and common force levels) than females (Onambélé, et al., 2007a). The findings of the current and previous studies may not be comparable as participants in the current study were full-time athletes while they were only recreational athletes in previous studies. Furthermore, in previous comparisons the sample populations were not participating specifically in jumping activities that are more specific to loading the knee extensor mechanism. Not accounting for the menstrual cycle at the time of testing of female athletes was not seen to interfere with results in the current study, as it has been demonstrated that PT mechanical properties were unaffected due to fluctuating estrogen levels (Burgess, et al., 2009c).

In the current study it was demonstrated that the normal and healthy PT maintains a relatively stable range of mechanical behaviour in support of previous findings. It does not appear that activity level alters the loading behaviour of the PT significantly across different populations and that the PT mechanical properties of elite level athletes have a similar pattern of mechanical behaviour to the general population. However, within a distinct subset of volleyball players there may be changing patterns of PT loading behaviour related to task-specific loading strategy, with a difference in PT stiffness based on the preferred landing leg.

6.5.1 Patellar Tendinopathy

Mechanical properties were similar between patellar tendons with and without patellar tendinopathy. This finding supports a study on PT mechanical properties between tendons classified with patellar tendinopathy compared to unaffected tendons prior to an intervention strategy, which also demonstrated no differences between groups (Kongsgaard, et al., 2009). The PT mechanical properties were maintained, following a heavy slow resistance strengthening program over 12 weeks, even though fibril density improved in the patellar tendinopathy group (Kongsgaard, et al., 2010). In contrast, at the ankle, the Achilles tendon affected by tendinopathy was reported to have lower stiffness and elastic modulus than normal Achilles tendons (Arya, et al., 2010). However, this recent study may be subject to confounding variables with the affected population (Achilles tendinopathy) having a trend for lower Achilles tendon force ($p = 0.073$) and greater body mass ($p = 0.06$) than the control group. Thus, differences in stiffness (force/CSA) not normalised for body mass may increase the difference between groups. The current study on PT mechanical properties is difficult to compare to findings of Arya et al. (2010) as the PT and Achilles are subject to different loads (Finni, et al., 2001; Kovács, et al., 1999; Kubo, et al., 2006; Maganaris, Narici, & Maffulli, 2008) and may differ morphologically (Kongsgaard, et al., 2010; Rees, Wilson, & Wolman, 2006; Tan & Chan, 2008). The current study was on elite level athletes while the activity level of the recruits in the study by Arya et al. (2010) is not reported.

It is important to note the potential differences in PT mechanical properties between individuals with and without patellar tendinopathy. In the unaffected men's volleyball group only 1 of 36 patellar tendons had a PT strain value less than two standard deviations from the mean. This was the same in the men's basketball group with 1 of 26 patellar tendons with a PT strain value less than two standard deviations from the mean of the group. In contrast, 4 of 7 patellar tendons with tendinopathy bordered the lower 95% CI in men's volleyball players and 2 of 5 patellar tendons with tendinopathy bordered the upper 95% CI in men's basketball players. Proportionately, more abnormal tendons have strain that is greater or less than the 95% CI for a comparative group with normal patellar tendons in both sporting groups. Therefore, investigation of differences in PT strain that may be outside the normal range may be warranted to identify risk or presence of patellar tendinopathy.

All other mechanical properties were similar between normal patellar tendons and tendons with patellar tendinopathy. The loading behaviour of all patellar tendons with tendinopathy remained within one standard deviation from the mean of the comparative group, when described by the value for elastic modulus that is the ratio of force normalised to CSA and PT deformation normalised to PT length at rest.

6.5.1 Patellar Tendon Cross-Sectional Area

Patellar tendon CSA was symmetrical between left and right sides at all three sites along the length of the PT. Although there was a trend in jumping sport groups for the proximal PT CSA to be greater than mid or distal sites, the only statistical difference detected was a greater proximal than mid PT CSA in the swimming group. This is in contrast to previous reports that have demonstrated greater distal PT CSA compared to the mid or proximal sites (Couppé, et al., 2008; Kongsgaard, et al., 2007; Westh, et al., 2008) possibly due to an accumulation of aggrecan that are shown to be present in areas where the tendon is susceptible to bending and twisting (Vogel, 2004). These previous MRI studies have described capturing images that are perpendicular to the PT however, it is unclear how this can be achieved, as in a neutral knee position in an MRI coil, the PT may lie at an angle oblique to the horizontal plane and hence MRI images may not be directly perpendicular to the PT. There is also a difference in the individuals tested with the current study recruiting full-time athletes compared to the previous studies that have recruited either non-jumping elite athletes (fencers and badminton players) (Couppé, et al., 2008), recreational runners and non-runners (Westh, et al., 2008) and untrained men (Kongsgaard, et al., 2007). Thus, differences in CSA along the PT may be specific to the exposure to the type of activity rather than have a similar pattern across different athlete populations. The determination of imaging changes in the PT may support the hypothesis of differential or non-uniform strain compared to a PT with no imaging changes (Sandercock, 2005). Even though differences in PT CSA between abnormal and normal tendons were not demonstrated in this study, future studies that are specific to measuring only collagen fibres that transmit tensile load, rather than the macroscopic appearance of the whole PT, may more specifically demonstrate changes in stress and therefore elastic modulus of the PT (Magnusson, et al., 2003b). The current and previous studies support the notion that that there is activity-specific adaptation of the PT CSA in athletes that may be a consequence of how the PT may be loaded.

6.5.2 Strengths and Limitations

Sampling a relatively homogenous group of elite level jumping and non-jumping athletes that train on a full-time basis, as live-in athletes, at an elite level training facility limited confounding factors that may impact on tendon loading behaviours. Participants consisted of elite level (national level) jumping athletes in volleyball and basketball that trained full-time (~30 hours/week) over the course of a calendar year. Participants were exposed to a number of modes of training in addition to ADL, including leisure and recreational activities. The loading history on the legs is impossible to control, however, this study aimed to sample a relatively homogenous group of elite level jumping athletes where these parameters were relatively controlled as athletes lived on the same site as where they trained.

The pilot study (chapter 5) supports the use of the method of measure of PT elongation. In light of evidence presented in this study and from the previous studies, it is vital to account for the complete PT elongation during a ramp MVIC. It has been discussed (6.5.1) that EMG does not provide an adequate method to assess co-contraction during knee extensor contraction, however, the inability to account for antagonist contribution to PT elongation is a limitation of this study. Investigation of athletes that are generally greater in stature and more susceptible to patellar tendinopathy than the general population necessitated a modified technique for measurement of PT elongation specific to the posterior portion of the PT.

A limitation in this study is the small sample size in each group. This affects the ability to testing the hypothesis that sport affects the preferred landing leg and may be tested in future investigations that examine the association between function and limb dominance. Furthermore, the non-jumping group of swimmers had only 6 participants and may not allow valid comparison between groups for PT mechanical properties of jumping athletes that were greater in number. Thus, comparisons between groups may lack external validity.

6.5.3 Summary and Conclusion

The mechanical behaviour of the human PT under load is similar for elite athletes that participate in different jumping and non-jumping sports and for the PT that is affected by tendinopathy. There is a large range of tensile force to which the PT may be exposed. It is clear that the normal and healthy PT is able to accommodate and cope with these tensile forces with potential for adaptation and change in CSA and strain such that the mechanical behaviour under load (modulus) remains similar between individuals. Patellar tendon tissue adaptation may be based on task-specific loading strategies as evidenced in elite level men's volleyball players in this study

7 Chapter Seven: A Prospective Study of *In Vivo* Patellar Tendon Mechanical Properties in Elite Athletes

7.1 Chapter Aims

This chapter aims to:

- compare the patellar tendon mechanical properties over a competitive season in elite level jumping athletes.
- describe patellar tendon mechanical properties of pathological patellar tendons over a competitive season in elite level jumping athletes.

7.2 Introduction

Patellar tendon mechanical properties are shown to change with training, unloading, disease and ageing. The evidence is conflicting as to whether there is a change in mechanical behaviour of the PT following exposure to strength training. Kongsgaard et al. (2010) demonstrated that PT mechanical properties remained similar following 12 weeks of knee extensor strength training for normal patellar tendons while there was a reduction in PT stiffness for tendons with patellar tendinopathy (Kongsgaard, et al., 2010). In contrast, neither slow heavy resistance nor eccentric exercise programs changed PT stiffness or elastic modulus of participants with patellar tendinopathy (Kongsgaard, et al., 2009).

Findings in two prospective studies on PT mechanical properties following strength training utilising leg exercises for 12 weeks in young men (Kongsgaard, et al., 2007) and 14 weeks in elderly recruits (N. D. Reeves, et al., 2003a), resulted in an increase in PT stiffness. Reeves et al. (2003) reported a decrease in PT strain in elderly recruits only, with a concomitant increase in elastic modulus (N. D. Reeves, et al., 2003a) however, no change in PT strain or modulus was demonstrated in young male recruits (Kongsgaard, et al., 2007). Similarly, in a group of young male recruits, isometric knee extensor training resulted in no significant changes in PT mechanical properties (Kubo, et al., 2006). During training and competition the intensive load and change in training load may lead to adaptations, however this has not been examined. Therefore, this study aimed to determine whether patellar tendon mechanical properties change after a season of training and competition.

The PT is susceptible to injury at the cellular level with significant disruption to tendon structure in some athletes participating in volleyball and basketball (Khan, Bonar, Desmond, Cook, Yong, Visentini, Fehrmann, Kiss, O'Brien, Harcourt, Dowling, O'Sullivan, Crichton, Tress, & Wark, 1996; Khan, Cook, Bonar, Harcourt, & Astrom, 1999; Khan, et al., 1998). In a population of recreationally active adults, patellar tendons affected by tendinopathy had reduced stiffness following strength training (Kongsgaard, et al., 2010) and exercise rehabilitation protocols (heavy slow or eccentric knee extensor exercise) for patellar tendinopathy (Kongsgaard, et al., 2009). The PT mechanical properties for tendons with tendinopathy were shown to behave similarly under load than those of a comparative group with normal patellar tendons bilaterally in chapter 6. It remains unknown how PT mechanical properties may change in elite level athletes with patellar tendinopathy exposed to one year of full-time training and competition.

Animal studies with long periods of training support a trend of increased tensile strength to tendon failure in rabbits that did 40 weeks of running training (Viidik, 1967, 1969) and swine extensors following 12 months of running training (Woo, Ritter, Amiel, Sanders, Gomez, Kuei, Garfin, & Akeson, 1980). Unfortunately it is difficult to do these studies in humans as tensile strength measures are generally performed *ex vivo* to tendon failure (Buchanan & Marsh, 2002).

Tendon hypertrophy is also altered by strengthening regimens in both humans and animals. Tendon hypertrophy has been reported following a 12 week bout of either light or heavy knee extensor training with an increase in CSA of mean 7% (SEM) (3) and 6% (3) respectively, at the proximal PT site (Kongsgaard, et al., 2007). Equine tendons (Birch, McLaughlin, Smith, & Goodship, 1999) and swine extensor tendons also increased in CSA following long-term training (Woo, et al., 1980), however, this was not found in the Achilles tendon of the guinea fowl (Buchanan & Marsh, 2001) or mice (Michna & Hartmann, 1989). The pattern of tendon hypertrophy is not consistent in animal studies as it varies between animals and is dependent on tendon function (high-stress flexors and low-stress extensors in quadrupeds) and the age of the animals tested (Buchanan, et al., 2002).

The association between exercise and tendon hypertrophy was supported by the finding of increased Achilles tendon CSA in runners compared to non-runners (Rosager, et al., 2002) and greater PT CSA in the lead leg of fencers and badminton players with knee extensor strength asymmetry (Couppé, et al., 2008). Similarly, greater PT CSA was found in the PT of the donor leg compared to the unaffected leg 1 to 10 years following anterior cruciate ligament reconstruction surgery and attributed to scar tissue formation with similar PT stiffness between sides (N. R. Reeves, et al., 2009). In contrast, PT hypertrophy has not been shown to occur in young adults (Kjaer, 2007) with only moderate increases in the elderly (N. D. Reeves, et al., 2003a). At the Achilles tendon, a prospective study found no significant change in Achilles CSA following 9 months of running training (P. Hansen, Aagaard, Kjaer, Larsson, & Magnusson, 2003). Key differences across these studies that make it difficult to establish whether tendon hypertrophy is expected following chronic PT load are the duration between testing periods, the level of activity of the sample population and the age of the individuals tested. This study aimed to evaluate whether tendon hypertrophy was a feature of PT adaptation following long-term exposure to repeated-loading, as this has not been previously investigated in jumping athletes who are susceptible to 'overuse' injury of the PT (Lian, et al., 2005).

It is evident that PT mechanical properties may change due to exposure to quadriceps strength training, ageing, surgery (ACL graft harvest) and unloading. These findings suggest that PT mechanical properties are susceptible to adaptation However, this has not been empirically determined, prospectively, over a full sporting season in elite athletes who participate in intensive training and competition. The purpose of this investigation was to evaluate changes in PT mechanical properties in elite jumping athletes.

7.2.1 Hypotheses

The derived variables for the domain of PT mechanical properties are described in the methods section (7.3.4). This investigation tested the following hypotheses for differences in PT mechanical properties, PT force and cross-sectional area after one year of sports participation in elite athletes.

1. That there will be no change in PT mechanical properties, force and cross-sectional area after one year of sports participation.

2. That there will be no difference in PT mechanical properties, force and cross-sectional area between normal an abnormal tendons.

7.3 Methods

7.3.1 Study Design

This was a prospective cohort design. All participants included in this study participated in the cross-sectional study (chapter 6) of *in vivo* PT mechanical properties. The number of days between testing sessions for participants was a mean of 344 (SD) (95.3).

7.3.2 Participants

Thirteen male athletes from the Australian Institute of Sport athlete population volunteered for participation in this study (Table 7.3.1). Participants were full-time volleyball (n = 11) and basketball (n = 2) athletes. Inclusion and exclusion criteria are the same as in chapter 6 (section 6.3).

Table 7.3.1 Descriptive characteristics (mean (SD)) of the thirteen male jumping athletes tested a mean of 344 (SD) (95) days apart.

	Age (years)	Height (cm)	Weight (kg)
Session 1	17.9 (2.1)	200.8 (4.0)	90.7 (8.3)
Session 2	18.9 (2.4)	202.2 (3.7)	92.5 (6.6)

7.3.3 Clinical Examination

The method for clinical examination was identical to that described in chapter 6 (section 6.3).

7.3.4 Patellar Tendon Mechanical Properties

The method for measure of *in vivo* PT mechanical properties of normalised PT force, elongation, stiffness, strain, CSA, stress and elastic modulus was identical to that described in chapter 6 (section 6.3).

Mechanical work done was calculated as ‘area under the curve’ for the force-length relationship. Work was calculated for the 10% intervals from 0-10% up to 90-100%MVIC as the area under the curve in units for force (N) and elongation (mm). Each 10% interval was converted to a percentage of work done from the sum of all 10 areas that represented 100%

of work done. The sum of the areas from each 10% interval from 50 to 100%MVIC was calculated to represent the percentage of mechanical work done from mid-range to maximal effort during a ramped MVIC knee extensor contraction.

7.3.5 Data Processing

Data processing was identical as that described in chapter 6 (section 6.3).

7.3.6 Data Analyses

7.3.6.i Clinical Examination

Paired t-tests were performed comparing the participants' height and weight between sessions (PASW Statistics, version 18). Significance was accepted at $p < 0.05$. The observed power for each statistical test was calculated using G*Power 3.1.2 (available as freeware on http://www.psycho.uni-dusseldorf.de).

Descriptive data from testing sessions 1 and 2 are presented for the thirteen participants who were included in this study. Data collected during clinical examination (chapter 6, section 6.3) includes scores on the VISA questionnaires, two visual analogue scales and classification of each PT as normal, clinical tendinopathy, imaging tendinopathy or both clinical and imaging tendinopathy. Patellar tendon mechanical properties including normalised PT force, strain, stiffness, stress and elastic modulus are also presented for the group of participants with normal patellar tendon bilaterally and for each participant with patellar tendinopathy.

7.3.6.i Patellar Tendon Mechanical Properties

Only 9 participants that had normal patellar tendons bilaterally at both assessments were included in the statistical analysis for comparison between sessions and sides. A 2 x 2 ANOVA (2 sides and 2 testing sessions) was used to compare PT mechanical properties (force, elongation, stiffness, strain, stress, elastic modulus and percentage mechanical work done) between sides and testing sessions with Bonferroni correction made for multiple analyses and significance accepted at $p < 0.05$. If there was no significant interaction between side and session, the significant main effect(s) were identified and significant pairwise comparisons were reported.

When a significant 2-way interaction was detected, one factor (side or session) was removed and a 1-way ANOVA performed on each factor independently. Bonferroni correction as made for multiple analyses and significant pairwise comparisons reported.

For each side (left and right), a 2-way ANOVA (sessions (2 levels) and site (3 levels)) was used to compare PT CSA between sessions and sites with Bonferroni correction made for multiple analyses and significance accepted at $p < 0.05$. If there was no significant 2-way interaction, significant main effect(s) were identified and significant pairwise comparisons were reported with significance accepted at $p < 0.05$.

If a significant 2-way interaction was detected, a 1-way ANOVA was performed on each factor independently. Bonferroni correction was made for multiple analyses for each ANOVA performed and significance accepted at $p < 0.05$. Significant pairwise comparisons were reported.

All statistical analyses were performed using PASW Statistics, version 18. The observed power for each pairwise comparison was calculated using G*Power 3.1.2 (available as freeware on http://www.psycho.uni-dusseldorf.de).

7.4 Results

7.4.1 Clinical Examination

There was a statistically significant increase in height from session 1 to session 2 ($p = 0.024$) but not weight ($p = 0.104$) (Appendix O: Table O.1).

Nine of the 13 participants had a normal left and right PT on both testing occasions (Table 7.4.1). Of the four remaining participants, 3 had left sided patellar tendinopathy initially and all had a resolution of symptoms. On the right, 2 of 4 participants had clinical and imaging patellar tendinopathy that remained symptomatic on follow-up examination. A single participant who had a normal left and right PT initially had bilateral clinical and imaging patellar tendinopathy on follow-up examination.

Table 7.4.1 Clinical profile of participants across sessions 1 and 2.

Pareti-cipant	Sport	Session	Patellar Tendinopathy Classification		VISA		VAS (1)		VAS (2)	
			Left (L)	Right (R)	Left	Right	Left	Right	Left	Right
1	VB	1	Normal	Normal	98	98	63	12	55	10
		2	Normal	Normal	96	97	14	0	34	9
2	VB	1	Normal	Normal	87	100	38	0	0	3
		2	Normal	Normal	81	81	44	40	18	29
3	VB	1	Normal	Normal	76	91	8	5	3	0
		2	Normal	Normal	73	69	0	0	13	0
4	VB	1	Normal	Normal	96	97	5	3	0	0
		2	Normal	Normal	84	85	14	0	12	0
5	VB	1	Normal	Normal	94	63	16	71	5	47
		2	Normal	Norma	57	79	28	0	26	10
6	VB	1	Normal	Normal	75	100	64	0	0	8
		2	Normal	Normal	82	100	55	2	0	0
7	VB	1	Normal	Normal	94	91	17	4	5	13
		2	Normal	Normal	100	100	0	0	0	0
8	BB	1	Normal	Normal	95	100	18	2	10	0
		2	Normal	Normal	91	100	17	3	58	0
9	BB	1	Normal	Normal	97	97	17	11	2	5
		2	Normal	Normal	85	100	39	3	1	2
10	VB	1	Clinical & Imaging	Normal	42	53	71	17	25	27
		2	Imaging	Normal	76	70	0	0	13	71
11	VB	1	Clinical & Imaging	Clinical & Imaging	70	70	81	68	78	37
		2	Imaging	Clinical & Imaging	80	92	8	19	20	25
12	VB	1	Clinical	Clinical & Imaging	86	86	23	26	0	0
		2	Normal	Clinical & Imaging	100	77	0	26	0	3
13	VB	1	Normal	Normal	89	75	33	40	7	31
		2	Clinical & Imaging	Clinical & Imaging	64	72	11	33	13	64

VB – Volleyball; BB – Basketball; VISA – Victorian Institute of Sport Assessment; VAS – Visual Analogue Scale; (1) - VAS recording pain during last one week of activity; (2) – VAS recording of pain during single leg decline squat

7.4.2 Patellar Tendon Mechanical Properties

Patellar tendon mechanical properties for the 9 participants with normal patellar tendons bilaterally in testing sessions 1 and 2 and the individual values for participants with patellar tendinopathy (participants 10 to 13) are reported in Table 7.4.2.

Table 7.4.2 Patellar tendon mechanical properties (mean (SD)) for the group with normal patellar tendons bilaterally (n = 9) and participants with patellar tendinopathy across two testing sessions on the left and right sides.

		Side	Force ($N/BM^{2/3}$)		Proximal CSA (mm^2)		Elongation (mm)		Stiffness ($kN.mm^{-1}$)		Strain (%)		Stress (MPa)		Modulus (GPa)	
			S1	S2	S1	S2	S1	S2	S1	S2	S1	S2	S1	S2	S1	S2
$Normal_{(n=9)}$		L	197.8	204.8	181.2	204.7	6.6	5.4	3.0	2.1	12.6	11.2	22.1	21.1	0.9	0.5
			(59.7)	(64.2)	(23.8)	(20.5)	(2.1)	(1.1)	(1.5)	(0.9)	(4.2)	(2.2)	(5.9)	(7.3)	(0.5)	(0.2)
		R	195.6	201.8	187.9	196.2	6.2	5.1	2.3	1.9	12.0	10.3	21.3	20.7	0.6	0.5
			(62.2)	(57.7)	(23.9)	(24.1)	(2.4)	(1.0)	(1.1)	(0.9)	(4.0)	(1.7)	(6.4)	(6.7)	(0.3)	(0.2)
Participant	10	L	200.5	184.1	209.9	183.2	8.6	7.1	0.87	1.1	16.1	13.9	19.8	20.5	0.2	0.3
		R	194.1	177.9	182.4	199.4	6.0	6.4	2.6	0.9	10.8	13.2	22.0	18.2	0.8	0.2
	11	L	96.4	130.3	199.1	226.5	3.7	4.0	2.6	1.32	6.6	8.7	10.5	12.7	0.7	0.3
		R	40.1	86.1	236.6	269.9	2.9	3.0	2.8	1.8	5.7	6.6	3.7	18.2	0.6	0.3
	12	L	129.0	219.0	196.0	208.6	3.9	4.8	1.6	1.5	6.7	10.3	12.9	21.4	0.5	0.3
		R	251.5	187.0	209.2	193.7	2.7	4.9	3.6	1.3	5.1	9.6	23.6	19.7	0.9	0.4
	13	L	162.4	98.1	213.3	204.3	5.4	2.9	1.5	1.5	12.5	7.3	16.0	10.1	0.3	0.3
		R	168.6	125.3	225.3	213.3	6.7	5.2	5.0	0.7	16.0	11.1	15.7	12.4	0.9	0.2

S1 – Session 1; S2 – Session 2; BM – Normalised to Body Mass (kg)

There were no significant interactions ($p > 0.05$) or main effect for side ($p > 0.05$) or session ($p > 0.05$) for normalised PT force (Appendix O: Figure O.1), elongation (Appendix O: Figure O.2), strain (Appendix O: Figure O.3), or stress (Appendix O: Figure O.4).

7.4.2.i Patellar Tendon Stiffness

There was no significant interaction ($p > 0.05$) or main effect for session ($p > 0.05$) or side ($p > 0.05$) at any 10% interval from 20 to 40% MVIC or from 50 to 80% MVIC ($p > 0.05$) (Figure 7.4.1) (Appendix O: Table O.2). There was a significant interaction for PT stiffness calculated from 40 to 50% MVIC ($p = 0.047$) (Appendix O: Table O.2). There was significantly greater PT stiffness on the right compared to left side from 40 to 50% MVIC during session 1 ($p = 0.044$) (Appendix O: Table O.3). An example of a force-length curve for the left and right side of a single participant is shown in Figure 7.4.2.

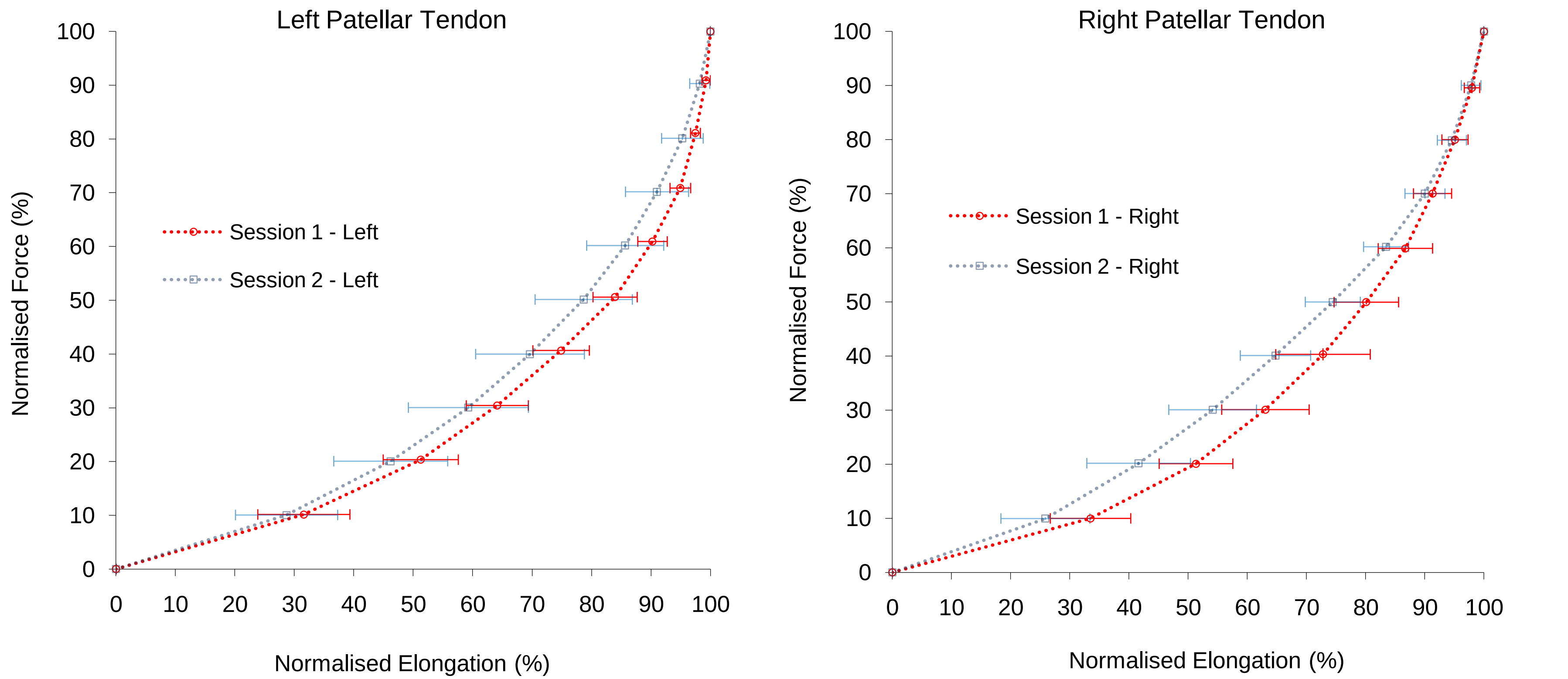


Figure 7.4.1 Normalised force-elongation curves (mean SD). Patellar tendon stiffness (slope at each 10% increment of force) is not significantly different between sessions 1 and 2 for either side. There is a trend for the force-length curve to 'shift' to the left (dotted red and blue lines) for the left and right patellar tendons.

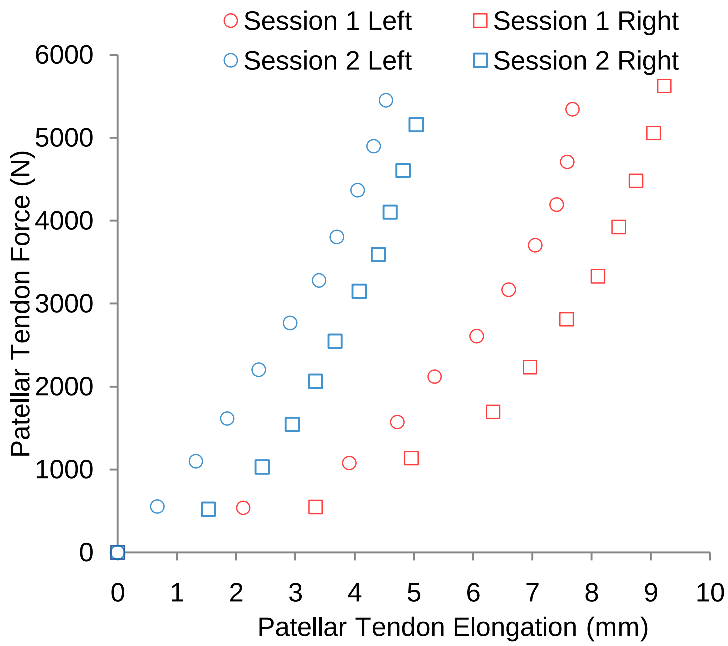


Figure 7.4.2 Patellar tendon force-length curve for a single participant (men's volleyball player) for left and right sides during each testing session from 0 to 100% maximal voluntary isometric contraction effort.

7.4.2.ii Patellar Tendon Elastic Modulus

There was significantly lower PT elastic modulus in session 2 compared to session 1 calculated from 70 to 80% MVIC ($p = 0.03$) (Figure 7.4.3) (Appendix O: Table O.5). An example of a stress-strain curve for the left and right sides of a single participant is shown in Figure 7.4.4. There was significantly greater PT elastic modulus on the right compared to left side when calculated from 40 to 50% MVIC ($p = 0.008$) (Appendix O: Table O.5).

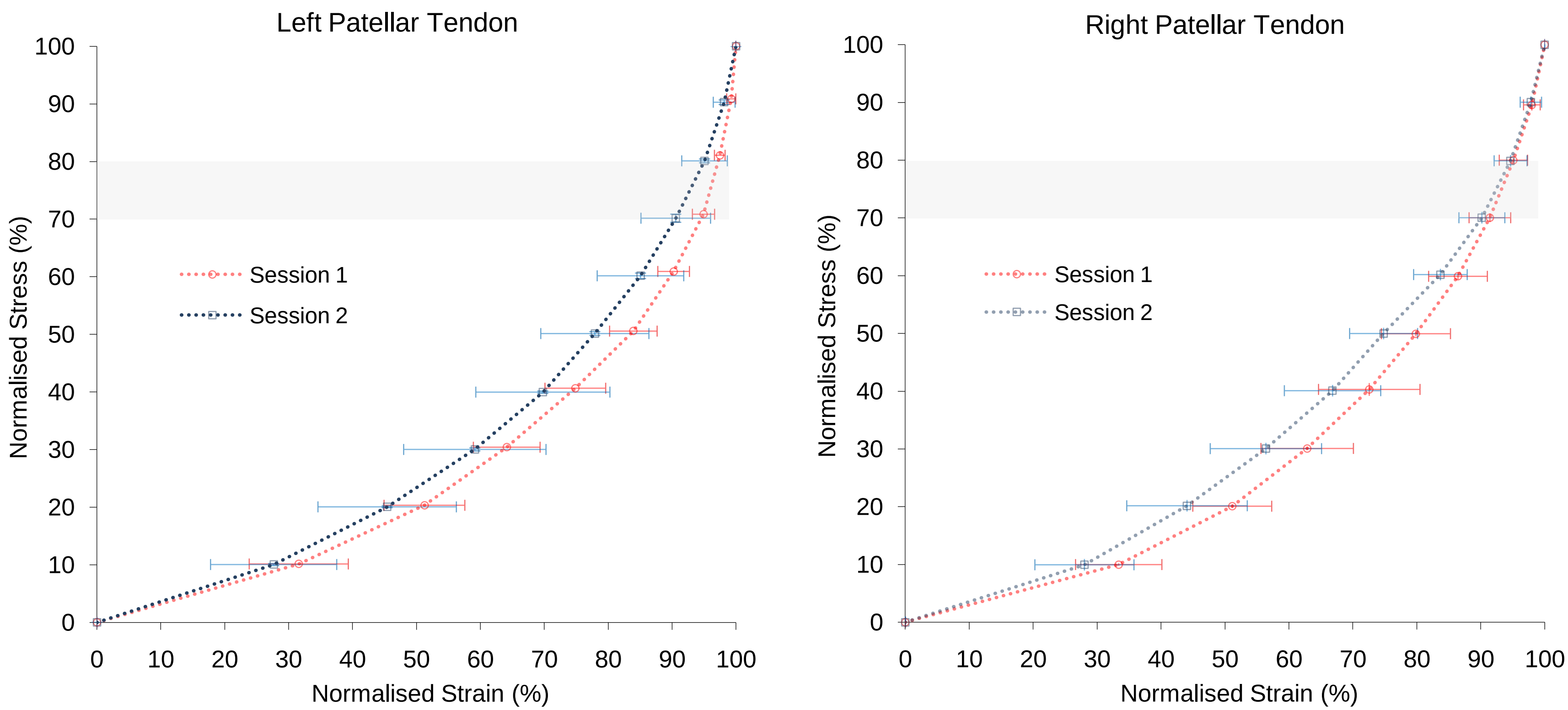


Figure 7.4.3 Normalised stress-strain curves (mean SD). Patellar tendon elastic modulus (slope at each 10% increment of force) is significantly lower during session 2 compared to session 1 ($p = 0.03$) at the 70 to 80% maximal voluntary isometric contraction effort increment (grey shaded areas) for the left and right patellar tendons.

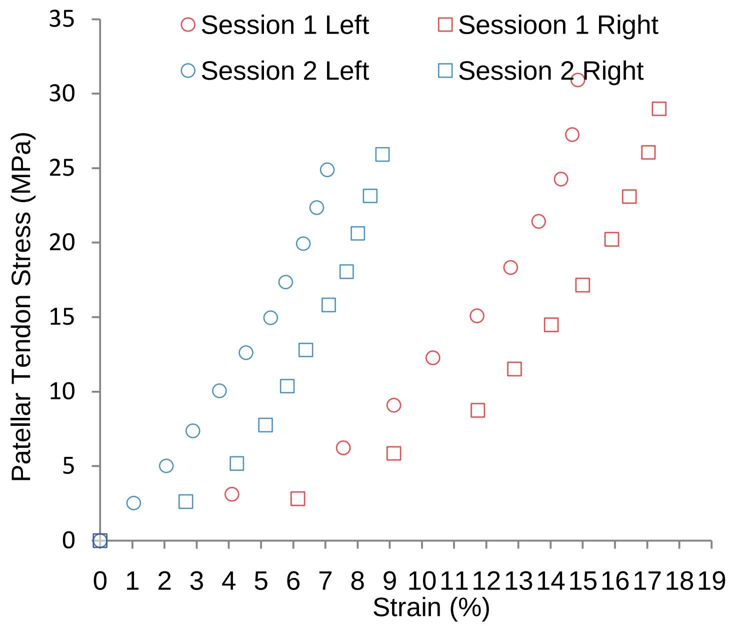


Figure 7.4.4 Patellar tendon stress-strain curve for a single participant (men's volleyball player) for the left and right sides during each testing session from 0 to 100% maximal voluntary isometric contraction effort.

<u>7.4.2.iii Patellar Tendon Work</u>

There was a significantly lower percentage of mechanical work done in session 1 compared to session 2 from 50 to 100% MVIC (p = 0.018) (Figure 7.4.5) (Appendix O: Figure O.5).

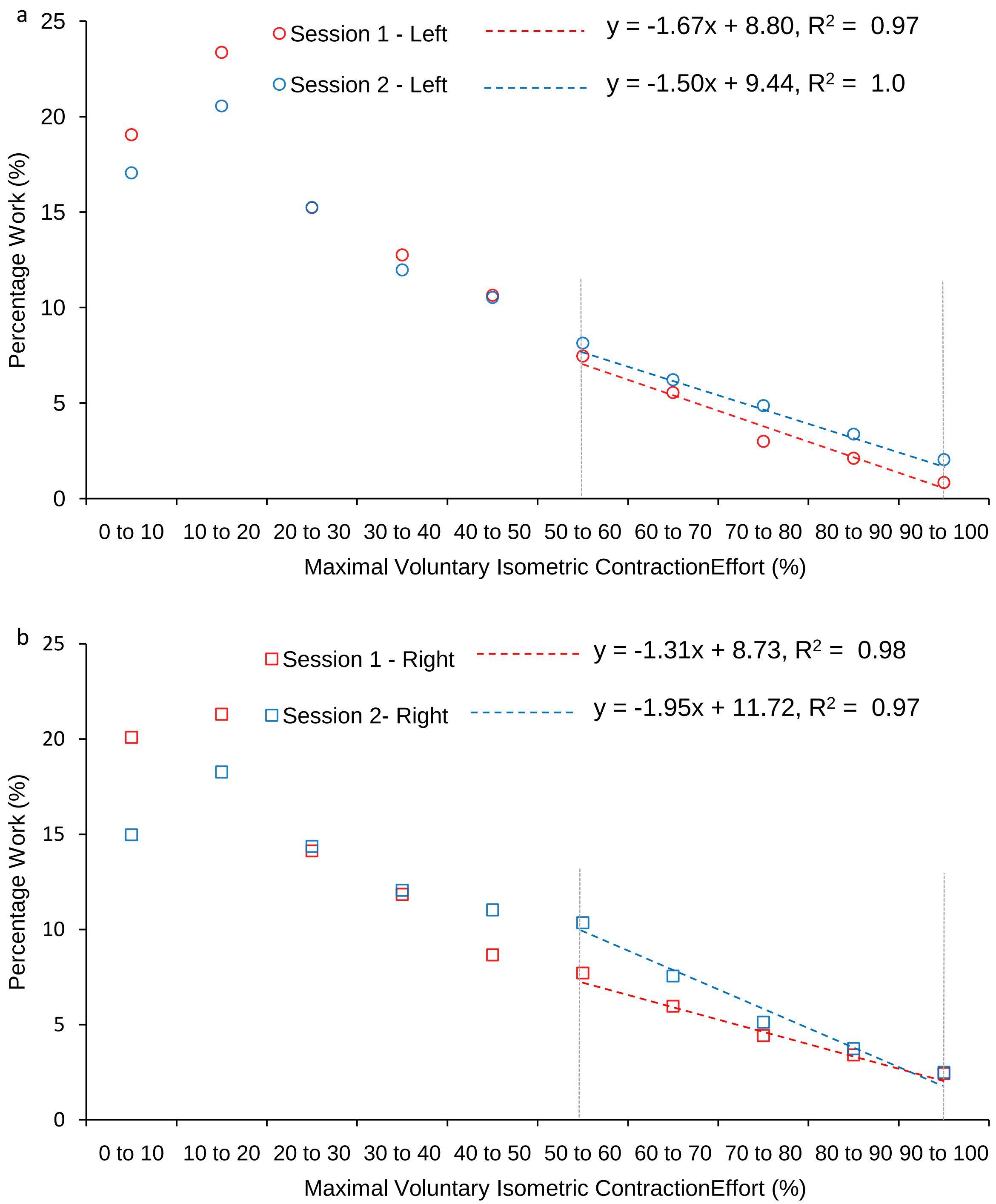


Figure 7.4.5 Percentage of patellar tendon work done for the left (a) and right (b) sides, calculated from 50% to 100% maximal voluntary isometric contraction (MVIC) effort (vertical dashed lines), was lower during session 1 than session 2 ($p = 0.018$). The linear regression line for work in this range (50-80% MVIC) in session 1 (red dashed line) and session 2 (blue dashed line) for both sides is strongly correlated with increasing effort from 50 to 100% MVIC effort ($R > 0.9$).

7.4.2.iv Patellar Tendon Cross-Sectional Area

Patellar tendon CSA was significantly greater at the proximal compared to the mid tendon site on the right side and no different between sessions for either the left ($p = 0.25$) or right ($p = 0.22$) (Figure 7.4.6) (Appendix O: Figure O.6 and Table O.8).

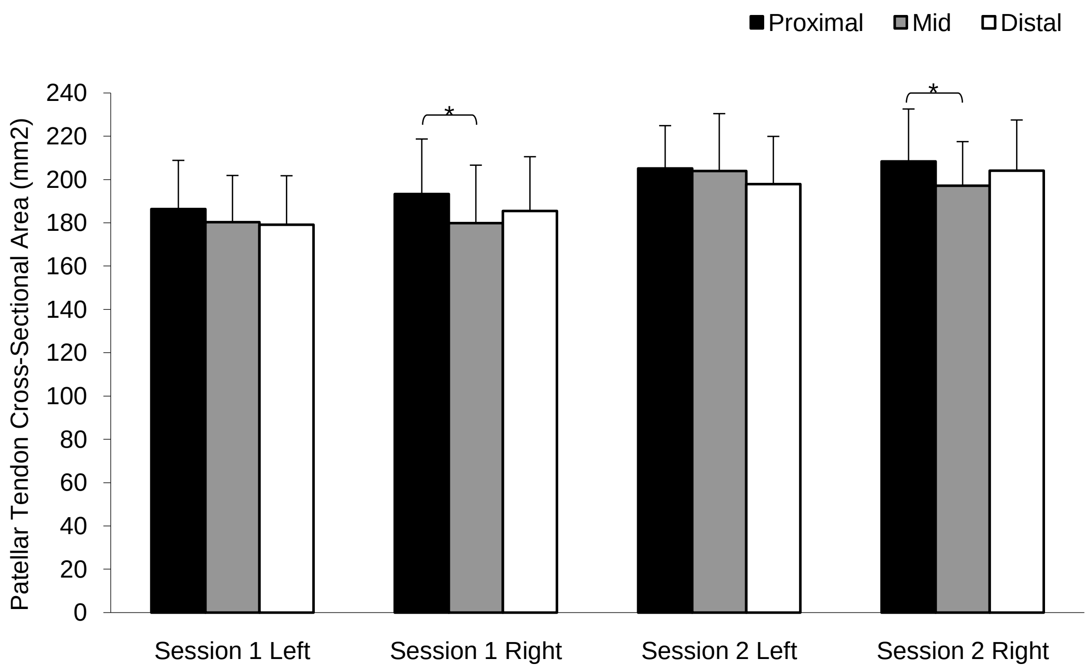


Figure 7.4.6 Patellar tendon (PT) cross-sectional area (mm^2) (mean SD) was significantly greater at the proximal compared to the mid PT site on the right side (*$p < 0.05$).

7.4.2.v Patellar Tendinopathy

The values for PT mechanical properties were generally within the 95% CI of the comparative group. One participant had a normal right PT with stiffness (60-80%MVIC) and work (50 to 100%MVIC) less than the 95% CI of the comparative group and went on to develop clinical and imaging tendinopathy. This same participant had a normal left PT with a mechanical work in session 1 that was greater than the 95% CI of the comparative group and went on to develop clinical and imaging tendinopathy. Another participant had a value for mechanical work of the right PT (clinical and imaging tendinopathy) that was less than the 95% CI of the comparative group that did not change its classification on follow-up.

7.5 Discussion

7.5.1 Clinical Examination

Only one participant developed bilateral clinical and imaging patellar tendinopathy that was initially normal on left and right sides. Three participants that initially had a symptomatic left PT had a resolution of symptoms. Two of these 3 participants also had right clinical and imaging patellar tendinopathy that had not resolved on follow-up examination.

7.5.2 Patellar Tendon Mechanical Properties

Derived variables were calculated for 9 of the 13 participants with normal patellar tendons bilaterally. The main findings in this study were an increase in percentage mechanical work done (from 50 to 100%MVIC effort) and lower elastic modulus (from 70 to 80% MVIC effort) of the PT in elite jumping athletes following a year of training and competition. The significant finding of a reduction in PT elastic modulus in the current study of mean 20% (SEM) (16.3) for full-time elite jumping athletes with normal tendons, was similar to the results of those following an exercise intervention for management of patellar tendinopathy, that reported a trend for reduction in elastic modulus for pathologic tendons (5% (54) to 13% (8)) (Kongsgaard, et al., 2009; Kongsgaard, et al., 2010).

In the current study, the important feature is that there is a trend for a reduction in the gradient of each curve at ranges from 20 to 80%MVIC for participants with normal patellar tendons bilaterally. This is in spite of a finding of no significant change in normalised PT force, strain or stress.

The current findings are in contrast to those on a young male population with healthy patellar tendons, who had an increase in PT elastic modulus (20%), stiffness (24%), PT CSA (7%) and muscle force (31%) following 9 weeks of dynamic knee extension strength training (Seynnes, et al., 2009) and increased PT stiffness following 12 weeks of heavy strength exercise (Kongsgaard, et al., 2007). In contrast, investigation of a similar population detected no change in PT mechanical properties following 12 weeks of isometric knee extensor strength training (Kubo, et al., 2006). From these previous studies and the contrasting findings of the current study, it is evident that the change in mechanical behaviour of the PT under load may adapt differently based on the loading history of the sample population as the current findings on elite level jumping athletes with normal patellar tendons are similar to the findings on participants with patellar tendinopathy and contrast to participants who are untrained and have healthy patellar tendons.

The sample population, which included elite level volleyball and basketball athletes, are likely subject to greater loads than the recruits in other studies who have primarily been untrained or recreationally active. This factor influences the loading history of the PT prior to testing, as the elite level athletes in the current study were participating for at least one year as full-time athletes prior to initial assessment and would have participated in their sport at a high level of competition and training prior to furthering their careers as a full-time athletes. Therefore, the PT would have been exposed to a stimulus great enough to influence its adaptation prior to initial assessment.

A further difference that makes it difficult to compare the current findings to previously reported results was the time between measures that have been taken up to 14 weeks apart (Kongsgaard, et al., 2009; Kongsgaard, et al., 2010; Kongsgaard, et al., 2007; Kubo, et al., 2006; N. D. Reeves, et al., 2004; Seynnes, et al., 2009). This was the first study that has measured elite level jumping athletes prospectively approximately one year apart. It is considered that there are differences in tendon adaptation that are partly dependent on the loading behaviour and the health of the tendon (Andersson, Eliasson, & Aspenberg, 2009; Buchanan, et al., 2002; Dahlgren, Mohammed, & Nixoon, 2005; Kjaer, Langberg, Miller, Boushel, Crameri, Koskinen, Heinemeier, LOlesen, Døssing, Hansen, Pedersen, Rennie, & Magnusson, 2005; Kjær, Magnusson, Krogsgaard, Bojsen Møller, Olesen, Heinemeier, Hansen, Haraldsson, Koskinen, Esmarck, & Langberg, 2006). The current study did not aim to assess the influence of an intervention of either strengthening or treatment on the PT but rather to evaluate changes that may be observed in PT mechanical properties of elite jumping athletes that are subject to all variables that may influence tissue loading (training, competition and activities of daily living) and health (injury and treatment) of the PT.

The level of skeletal maturity influences PT mechanical properties as demonstrated in comparison of young adult men and women 27-28 years of age, who have greater PT stiffness and elastic modulus than boys and girls (8 to 10 years of age) when compared at a common force level (O'Brien, et al., 2010). In the current study there was a significant yet small (0.8%) increase in height across testing sessions. This was unlikely to contribute significantly to the observed differences in PT mechanical properties as all participants were considered to be at the end or nearing the end of any significant growth. Reeves et al. (2003) reported an increase in PT elastic modulus and stiffness in an elderly population that was primarily due to a reduction in PT elongation and strain with no significant change in PT force (N. D. Reeves, et al., 2004). The magnitude of reduction in PT strain in the elderly (9.9% to 5.9%) (N. D. Reeves, et al., 2003a) was not found in studies in younger populations (Kongsgaard, et al., 2009; Kongsgaard, et al., 2010; Kongsgaard, et al., 2007; Seynnes, et al., 2009) or in the current study. Therefore, it was unlikely that age or skeletal maturity significantly influenced the changes observed in PT mechanical properties in the current study.

In the previous chapter (chapter 6), PT mechanical properties were shown to be similar between participants with normal tendons compared to those affected with patellar tendinopathy similar to previous studies (Kongsgaard, et al., 2009; Kongsgaard, et al., 2010). The current finding of a reduction in PT elastic modulus and greater percentage mechanical work done may indicate that a change in PT mechanical properties was related to the loading behaviour. Patellar tendinopathy, described as an overuse condition, was rehabilitated through exercise that also reduced stiffness and elastic modulus (Kongsgaard, et al., 2009; Kongsgaard, et al., 2010). The full-time jumping athletes in the current study, that are ‘overusing’ their PT compared to the general population, were found to have the same direction of change in mechanical behaviour of the PT under similar loading conditions. This supports the hypothesis that patellar tendinopathy may be a part of an adaptive process of the PT when over-loaded in the case of jumping athletes (Hamilton, et al., 2004).

The relatively greater period of exposure to training of elite and non-elite level jumping athletes may also be a conditioning response over time that increases the tendon’s ability to withstand tensile stress. This concept of ‘fatigability of tendons’ was demonstrated when comparing tendons from sheep of different maturity, whereby older sheep had a greater time to rupture and high stressed tendons (plantaris) were more resistant to tensile stress than low-stress (extensor digitorum lateralis) tendons (Pike, Ker, & Alexander, 2000). In humans the high-stress tendons are the lower limb extensors including the PT. As athletes will have greater exposure to high-stress tendon loading compared to untrained individuals, their relative resistance to tensile stress may be greater. The finding of a greater percentage of mechanical work done from 50 to 100% MVIC effort may be observed as the increased resistance to loading behaviour following a full year of training and competition. This may explain the difference in the direction of change for stiffness and elastic modulus that is reduced in athletes and individuals with patellar tendinopathy (Kongsgaard, et al., 2010) and increased in untrained individuals following strength training (Kongsgaard, et al., 2007; N. D. Reeves, et al., 2003a; Seynnes, et al., 2009).

Reasons for calculating the gradient at a near-maximal effort were reported in chapter 5 and briefly described here. As there was no PT elongation detected in some trials from 80 to 100% effort there is a division by zero for stiffness (force/elongation) and elastic modulus (stress/strain). There were no missing values for PT elongation from 70 to 80% MVIC efforts. Furthermore, the measure of stiffness and elastic modulus at near-maximal effort may be more representative of effort during sport and ADL considering volleyball and basketball do not always require maximal effort in jumping.

The calculation of mechanical work done by the PT from the range of MVIC effort from 50 to 100% effort was chosen as representative of a range of forces that participants would use in sports such as basketball and volleyball, while avoiding the lower efforts where the elongation of the PT may be due to creep (chapter 6). The relationship of mechanical work done to force (%MVIC effort) is shown to be strong ($R > 0.9$) and linear demonstrating that this was representative of the linear part of the force-length curve. Furthermore, including efforts >80% MVIC for the calculation of work (product of force and elongation), even if there was no PT elongation, allowed for normalisation of work values as a percentage of total work done.

7.5.3 Patellar Tendon Cross-Sectional Area

The current findings of no significant increase in PT CSA between sessions support the findings of no PT hypertrophy following quadriceps strength training (Kubo, Kanehisa, & Fukunaga, 2002; Kubo, et al., 2006; N. D. Reeves, et al., 2003a). However, heavy and light strength training protocols resulted in statistically significant increases of mean 6% (SEM)

(3) and 7% (3) respectively, in PT CSA for untrained males (Kongsgaard, et al., 2007) and increases from 103mm^2 (1) to 107mm^2 (1). These increases were similar to current study with an increase of mean 12.6% (SEM) (5.0) (proximal), 5.2% (10.8) (mid) and 4.6% (10.8) (distal) in PT CSA. This indicates that the short-term increase in PT CSA following strength training is similar to those following one year of training and competition. However, the studies may not be comparable as the training history of participants is different between sample populations.

Considering the elite level of athletes included in this study, all participants were participating at a high level of training and competition prior to inclusion in this study. Therefore, they would have had continuous exposure to relatively high PT loading behaviours compared to the general population. This supports the finding of greater PT CSA at both baseline and follow-up in the current cohort ranging from mean 178mm^2 (SEM) (9) to 205mm^2 (7) compared to other studies on males that report values of PT CSA ranging from 73mm^2 (20) (untrained) (Kubo, et al., 2006) to 163mm^2 (12) (recreational athletes) (P. Hansen, et al., 2006). The greater PT CSA found in this study compared to previous studies may represent PT hypertrophy specific to elite level jumping athletes as the greater values were present at both initial and subsequent assessments. Or, the greater measure of PT CSA may be due to the greater size of athletes in the current study compared to recruits in other studies. The current finding supports the concept that only prolonged training will lead to a net synthesis of collagen (Kjær, et al., 2006) that in these jumping athletes is highlighted by large PT CSA values. However, across all human studies it has yet to be definitively determined whether the increases in CSA are due to collagen or ground substance proliferation, or both.

The consistently greater PT CSA (183 to 270mm^2) across participants with or without patellar tendinopathy may represent an adaptation to full-time training and competition. As demonstrated previously (chapter 6) PT CSA in pathological tendons was not shown to be outside the 95%CI from the mean for the population of jumping athletes with normal patellar tendons bilaterally. Although the presence of hypoechoic areas and tendon thickening may be present in athletes diagnosed with patellar tendinopathy (Cook, et al., 2001a; Cook, et al., 2000b; Cook, et al., 2004b; Gemignani, et al., 2008; Gisslèn, et al., 2005; Peace, et al., 2006) a difference in PT CSA between affected and unaffected patellar tendons may not be sensitive or representative of a significant change in tendon morphology in elite level athletes that appear to have greater PT CSA than the general population. In the current study only one participant with patellar tendinopathy had PT CSA greater than the 95% CI of the comparative group. This participant had a right proximal PT CSA greater than the comparative group in both sessions, with the mid and distal CSA values only greater in session 2. Therefore, it may be considered that PT hypertrophy was maintained over the course of the year rather than causative of an enlarged PT in this participant. The finding that all other participants with patellar tendinopathy had PT CSA within the 95% CI of the comparative group implicates a change in PT tissue morphology at the fibrillar level as responsible for changes in mechanical behaviour rather than CSA (Magnusson, et al., 2003b).

A finding of lower fibril density and greater fibril area in patellar tendons affected with patellar tendinopathy was demonstrated in individuals prior to an exercise program (Kongsgaard, et al., 2010). The levels of fibril density and fibril area returned to levels close to those of the control group, following 12 weeks of heavy, slow exercise training that also reduced PT stiffness by a mean of 9% (SEM) (6) (Kongsgaard, et al., 2010). Unfortunately it is difficult to ascertain whether there is a change in fibril morphology following long-term exposure to exercise, as the bulk of the studies pertain to changes induced following relatively short periods of exercise of up to 10 weeks (Buchanan, et al., 2002).

In patellar tendinopathy there is myxomatous tissue in place of tight parallel collagen fibres (Khan, et al., 1999; Khan, et al., 1998). Hypoechoic regions on ultrasound imaging that correlate with the presence of deleterious tissue changes may be observed in the presence or absence of pain (Cook, et al., 1998; Khan, Cook, Kiss, Visentini, Fehrmann, Harcourt, Tress, & Wark, 1997; Khan, et al., 1998 ; Lian, Halen, Engebretsen, & Bahr, 1996). In the current study of the 26 tendons measured one year apart, 21 were classified as normal. However, of these 21 normal tendons, tendon thickening was noted in 9 tendons at baseline and in 6 tendons at follow-up. It has been demonstrated that the health of tendon tissue fluctuates and may be independant of the presence of pain or dysfunction (Cook, et al., 1998; Cook, et al., 2001a; Cook, et al., 2000b; Cook, et al., 2005; Gisslèn, et al., 2005) and for this reason in the current study, classification of tendinopathy was made both clinically (clinical patellar tendinopathy) and using imaging (imaging tendinopathy).

This change in appearance not only supports patellar tendinopathy to be part of an adaptive process (Hamilton, et al., 2004); it may also suggest the presence of non-uniform intra-tendinous stress along parallel collagen fibres that transmit load. It is well demonstrated that there is non-uniform strain along parallel tendon bundles of the PT as reported from *in vitro* studies (Almekinders, et al., 2002; Basso, et al., 2002; P. Hansen, et al., 2010; Haraldsson, et al., 2005), computer modeling (Lavagnino, et al., 2008) and *in vivo* (intra-tendinous optic fibre insertion technique) (Dillon, et al., 2008) studies. It may be hypothesised that the presence of abnormal intra-tendinous structure may alter the magnitude of stress at sites within the PT and that this may be linked to the pathogenesis of clinical patellar tendinopathy.

7.5.4 Strengths and Limitations

This study has the benefit of measuring a relatively homogenous group of elite level jumping athletes over a full-season and is unique in the literature to date on investigations of PT mechanical properties.

This study is limited in that there was no control group to allow comparison of changes to be attributed solely to the exposure of jumping athletes to training and competition. If a control group that participated in non-jumping elite level sport had demonstrated the same changes similar to the jumping group then it cannot be concluded that the changes were due to a training and competition in jumping sports. A further limitation is that training load and injury history is not known for participants over the course of the year. It is likely that there would be differences between individuals in the loading behaviour of the PT. Factors that will influence the loading behaviour over the course of the training and competitive season include differences in player position, differences in training and competition, injury history, treatment history (medication, exercise rehabilitation, rest, electrophysiological agents) and psychological factors such as motivation and the self-perception of wellness (Gabbett, et al., 2007; Le Bars, Gernigon, & Ninot, 2009; Mohamed, et al., 2009; Reilly, et al., 2009; Sheppard, et al., 2009; A. L. Smith, Balaguer, & Duda, 2006). By not being able to quantify the PT load and having a relatively small sample size, the observed changes may not be attributed directly to a training effect and may be due to one or more confounding variables. Some of these effects may be washed out considering the relatively long period of time between measures and that all athletes lived on site and had similar training loads and a common coaching and training staff to supervise sessions.

7.5.5 Summary and Conclusion

This study demonstrates that there was a change in mechanical behaviour under load of the PT at near-maximal efforts in elite jumping athletes. The finding of a reduced elastic modulus in the current study was similar to findings in recruits with patellar tendinopathy following strength training where there was a reduction in PT stiffness (Kongsgaard, et al., 2010) and a similar trend for reduced elastic modulus in recruits with patellar tendinopathy following strength training (Kongsgaard, et al., 2009). This suggests that normal patellar tendons of elite jumping athletes and those with patellar tendinopathy behave similarly following prolonged exposure to either specific knee extensor training, as demonstrated by Kongsgaard et al. (2009 and 2010) or from chronic exposure to training and competition in jumping sports at the elite level in the current study. Future studies may aim to determine whether a reduction in PT elastic modulus increases or decreases the risk of developing patellar tendinopathy.

This suggests that the mechanical behaviour of chronically over-loaded patellar tendons is unique and specific to the tendon loading history (Magnusson, Langberg, & Kjaer, 2010; Magnusson, et al., 2008) and the finding of greater percentage mechanical work done from mid-range to maximal effort may support the concept of a developed resistance to loading. In determining improved assessment and rehabilitation strategies for individuals that are either susceptible to or suffering from patellar tendinopathy, it is recommended that the history of tendon loading behaviour be determined before exercise or rehabilitation strategies are employed. This will affect the ability of the PT to either increase or reduce its ability to withstand deformation at different efforts of knee extensor contraction.

8 Chapter Eight: Discussion

8.1 Main Findings

8.1.1 Lower Limb Motor Control in Elite Athletes during Hopping

This is the first thesis, to the best knowledge of the author, to evaluate mechanical behaviour and muscle activation characteristics in elite level athletes using a number of different protocols. This thesis was specific to elite level athletes that participate in jumping sports as this population has a high prevalence of patellar tendinopathy (Lian, et al., 2005). This was in contrast to the majority of previous studies that have sampled recreationally active participants (Austin, et al., 2003; Farley, et al., 1999; Hobara, et al., 2007; McLachlan, et al., 2006; Padua, Arnold, Perrin, Carcia, & Granata, 2002; Padua, et al., 2006) or athletes participating in running (Hobara, et al., 2008; Hobara, et al., 2010b). Thus, an investigation of factors that may link lower limb mechanical behaviour to knee function was warranted to allow a greater understanding of how the motor control system may contribute to overuse injury at the knee.

The findings in chapter 3 support the existing evidence that the ankle joint is the major contributor for the modulation of lower limb mechanical behaviour during hopping on a flat surface at a self-selected hopping pace (Farley, et al., 1999). The current study supports the literature with the finding that changes at the ankle joint in stiffness (Farley, et al., 1999) and triceps surae muscle activation amplitude (Hobara, et al., 2007) modulate the lower limb mechanical behaviour when hopping conditions changed across the factors of surface (flat or decline), tasks (DL or AL) and pace (self-selected or forced) with a conservation of knee and hip stiffness. At the knee joint there was greater mechanical work done primarily as a result of greater knee joint excursion during the braking phase when hopping on a decline compared to flat surface. The current study further demonstrated that by constraining ankle joint function by hopping on a decline surface, there was an increased capacity of the knee joint to contribute to the maintenance of hopping. Previous findings determined that the knee joint was the main site for modulation of lower limb mechanical behaviour during maximal height hopping (Hobara, et al., 2009) and during a counter-movement jump following a drop jump (Horita, et al., 2002). Therefore, the current study supports the assertion that there is a complex interplay between knee and ankle mechanical behaviour that is specific to the loading of the lower limb. First, when there was no constraint to ankle function and the SSC was at a sub-maximal effort, the ankle joint was the primary site for modulation of lower limb mechanical behaviour. Second, when the ankle joint was constrained either via extrinsic factors such as a decline board as demonstrated in chapter 3, or via the demand of the lower limb to have to generate a specific output such as during a drop jump (Horita, Komi, Nicol, & Kyröläinen, 1996), horizontal component of a stop-jump task (Edwards, et al., 2010a) or maximal height hopping (Hobara, et al., 2009), then the knee joint will increase its ability to absorb mechanical work done during the braking phase.

Previous work has shown that knee stiffness, during a single-leg hopping task at a self-selected hopping pace, to be minimised relative to leg stiffness comparing participants with and without Achilles tendinopathy, with no changes in ankle stiffness (Arya, et al., 2006). When comparing participants with and without patellar tendinopathy, the relative contribution of the knee to total support moment during a single-leg hopping task at self-selected and paced hopping (1.66Hz) was reduced, with no difference in contribution by the ankle between groups (Souza, Arya, Pollard, Salem, & Kulig, 2008). These two studies demonstrate that when ankle joint function was not constrained and at sub-maximal efforts, changes in knee joint mechanical behaviour enable maintenance of steady state hopping in participants with either ankle or knee overuse injury. The knee joint was also shown to modulate lower limb mechanical behaviour in a variety of landing tasks that either limit the capacity of the ankle to contribute to absorbing kinetic energy by being near-maximal efforts or due to the task not allowing optimal storage of elastic energy in the calf complex (Bisseling, et al., 2007; Bonnard, et al., 1994; Edwards, et al., 2010b; Horita, et al., 2002; Hughes, et al., 2008; Kovács, et al., 1999; Marquez, et al., 2009; Orishimo, et al., 2006; D. P. Richards, et al., 2002; D. P. Richards, et al., 1996; Yeow, et al., 2010; S.-N. Zhang, et al., 2000). The current study supports the hypothesis that the SEC of the knee extensors have to adapt in jumping athletes who are chronically exposed to sub-maximal and maximal jumping efforts.

8.2.2 Patellar Tendon Mechanical Properties in Elite Athletes

This is the first thesis, to the best knowledge of the author, to investigate PT mechanical properties in elite level jumping athletes. The current findings of increased mechanical work and reduced elastic modulus following a year of training and competition in elite level jumping athletes is similar to the changes observed in a previous study that investigated the mechanical behaviour of patellar tendons following exercise rehabilitation for tendinopathy, which had a trend for reduced elastic modulus (Kongsgaard, et al., 2009; Kongsgaard, et al., 2010). The literature also reports changes in PT mechanical behaviour following strengthening, with increased elastic modulus, stiffness and CSA in young healthy populations (Kongsgaard, et al., 2007; Seynnes, et al., 2009). The findings from these previous studies and from the current study suggest that PT mechanical behaviour may be dependent on the loading profile of each participant rather than the type of sports participation. The current study also supports the hypothesis that chronic loading of the PT in elite level jumping athletes may lead to similar changes in PT mechanical properties as observed following rehabilitation of patellar tendinopathy. This implies that the presence or absence of pain at the site of the PT may not be discriminated by a difference in PT mechanical properties. Importantly, the differences in measures of PT mechanical properties in the current thesis, whether the PT was chronically loaded or pathologic, are considered a resultant effect across the whole PT length and cannot infer any localised changes in mechanical properties to a segment of the PT. This thesis adds to the evidence that the presence or absence of tendinopathy was independant of the patellar tendon's capacity to withstand tensile load (Hamilton, et al., 2004).

The finding from this thesis shows similar PT mechanical behaviour across elite level jumping and non-jumping athletes, specifically for elastic modulus, which demonstrates that the PT mechanical behaviour under load may be similar between individuals who participate in different types of sports. This is supported by two further observations. First, the finding of similar PT mechanical characteristics for patellar tendons with and without tendinopathy in the current thesis and previously (Kongsgaard, et al., 2009), may demonstrate that patellar tendons with a similar loading profile, regardless of health, have similar mechanical behaviour. Second, that increased stiffness and modulus following strengthening exercise of patellar tendons in healthy populations (Kongsgaard, et al., 2007; Seynnes, et al., 2009) was reversed in individuals with patellar tendinopathy or athletes that are chronically exposed to training and competition, as in the current thesis. Therefore, the PT tissue may aim to behave similarly between participants with the increase or decrease in stiffness or elastic modulus being dependent on an individual's loading history rather than a 'normal range' of values.

Although the increase or decrease in stiffness or elastic modulus for the PT may be associated with the loading history (Magnusson, et al., 2010), there was notable heterogeneity of values for PT mechanical properties between individuals. The findings of an association between the preferred landing leg and PT stiffness in men's volleyball players, suggests that the ability of the PT to adapt, is based on a change in PT loading behaviour, rather than an 'optimal' value of stiffness or elastic modulus. The difference in stiffness of the left PT between groups based on the preferred landing leg may be due to either; an increase in left PT stiffness of those who preferred to land on their right leg (potentially under-loaded), or a reduction in left PT stiffness of those who preferred to land on their left leg (potentially over-loaded). Therefore, detecting the relative increase or decrease in stiffness or elastic modulus will more accurately describe the adaptive process of the PT.

The mechanical behaviour may adapt based on the relative change with 'over' or 'under' loading of the PT and parallels the differences in change previously observed in groups of healthy recruits (Kongsgaard, et al., 2007; Seynnes, et al., 2009) compared to those with patellar tendinopathy (Kongsgaard, et al., 2009; Kongsgaard, et al., 2010) or athletes who are exposed to chronic over loading (chapter 7). The current findings highlight the direction of change in PT mechanical properties due to the relative overloading of the PT. First, the lower PT stiffness was demonstrated for the left PT of men's volleyball players who preferred to land on their left leg. This finding may be a consequence of players being right hand dominant during spiking and being forced to land on their left leg (Gabbard, et al., 1996a). Also, in players who prefer to land on their right leg, the left PT may be relatively stiffer, similar to the change observed in healthy volunteers following strength training (Kongsgaard, et al., 2007; Seynnes, et al., 2009). The lack of differences in PT stiffness on the right side serves as a control, highlighting, that even though both legs will experience the same number of loading episodes, whether during training, competition or ADL, the relative magnitude of loading during high intensity activity may be different between legs and act as the discerning factor to differentiate PT adaptive changes in mechanical behaviour. Second, the finding in the current thesis of lower elastic modulus following a season of training and competition may also be described as chronic overloading for the elite level jumping athletes assessed. In both cases, the common feature of relative PT overloading increased the compliance of the PT to tensile force. This is the first study (chapter 6) to suggest that laterality and forced landing preferences have a significant interaction. Therefore, one reason for diversity within the literature on PT mechanical properties may be a result of this phenomenon.

8.2 Strengths and Limitations

The strength of this thesis was to sample elite level athletes. Elite level jumping athletes are shown to have neural adaptation specific to their training (Masci, Vannozzi, Gizzi, Bellotti, & Felici, 2010) and are particularly susceptible to patellar tendinopathy that can be symptomatic for months or years (Cucurulo, Louis, Thanaut, & Franceschi, 2009). In athletes who need to train and compete year round, the ability to remain free from injury is paramount to their success. Therefore, this thesis sought to determine motor control behaviours and local tendon mechanical behaviour that may be explicit to elite level jumping athletes who are susceptible to patellar tendinopathy.

The within-subject study design for measure of lower limb motor control parameters utilised a number of different derived variables to describe the mechanical and muscle activation characteristics of the lower limb during a repeated SSC. The use of a decline board during hopping was a novel approach to preferentially increase the load of the knee joint. The use of an AL hopping task to test single-leg loading strategy, in addition to the commonly tested DL hopping task, was another novel approach to evaluate lower limb loading, as both strategies are used in sport and ADL. The spring-mass model is a commonly used method to model hopping mechanical behaviour. This thesis established that the spring-mass model was not accurate for measuring vertical leg stiffness for a number of hopping conditions (chapter 4). Hence, a first principles approach to calculating vertical leg stiffness was used. In addition to testing elite level athletes from a range of jumping and non-jumping sports, this thesis prospectively assessed the PT mechanical properties of elite level jumping athletes following a season of training and competition. This provided a unique insight into the change in PT mechanical properties in a group that is susceptible to patellar tendinopathy.

Measurement of *in vivo* PT mechanical properties was dependent on the accurate measurement of PT length. This thesis recorded proximal and distal PT segment lengths that was validated and shown to be reliable (chapter 5). This was particularly important in testing tall athletes, as the complete PT length was unable to be imaged on ultrasound during a single MVIC effort. The method of using an echoabsorptive marker and measure the proximal segment length has previously been used to monitor whether there is relative movement of the skin to the line cast by the hypoechoic skin marker (Maganaris, et al., 2004; N. D. Reeves, et al., 2003a). Trials in these previous studies were discarded if movement was noted. The current method (chapter 5) was developed to avoid the potential omission of trials and in fact determined that it was necessary to capture the resting length and elongation of each segment, then to sum the values for each segment. This method avoids the potential sources of error due to marker movement during contraction, marker movement during repositioning of the transducer to visualise different tendon segments and avoid under- or over-estimation of PT strain.

A limitation of the studies involving estimates of PT force was not measuring the contribution of antagonist muscle (hamstring) co-contraction. The ability to accurately quantify the load of the antagonist muscle is problematic and this is endemic in the literature (Grabiner, et al., 1992; Grabiner, et al., 1989; Kellis, 2003; Kellis, et al., 1999a; Kellis, et al., 2008; Kellis, et al., 2005; Kingma, et al., 2004; Krishnan, et al., 2010). Therefore, the force contribution, during a ramped maximal voluntary isometric contraction (MVIC) of the quadriceps femoris from 0 to 100% effort, was not collected during this study.

Although the proportion of contribution to PT elongation due to quadriceps and hamstrings activity may not be known, it was accepted that at the near maximal efforts (70 to 80% MVIC) the ‘net’ knee extension force (quadriceps femoris and hamstring activity) was responsible for the ‘net’ PT elongation. It was not the aim of these studies to determine how much of the PT elongation was due to the activity of the quadriceps femoris only. The data on knee moment was reliable and consistent with previous studies (Burgess, et al., 2009b; S. J. Pearson, et al., 2005; S. P. Pearson, et al., 2007). While it may be accepted that co-contraction affects the data, this may be systematically different between groups. Therefore the between sides and groups findings are valid in this model.

This thesis evaluated the inter-side differences in both dynamic and static testing procedures. The use of dual adjacent force plates and sEMG during hopping, allowed for calculation of joint related variables and muscle activation characteristics on both sides. The inter-side differences were evaluated for PT mechanical properties and allowed for evaluation of dominance related relationships between PT mechanical properties and function.

Limitations in this thesis are primarily related to the small sample size that may reduce the power of each study. This will also limit the comparison of results to other populations. The loading history of the legs was impossible to control; however, sampling a relatively homogenous group of athletes reduced the effect of this confounding variable. Another limitation was that there was no control group in the prospective evaluation of PT mechanical properties in elite jumping athletes. Thus, any significant differences may be due to a change in mechanical properties not attributed specifically to the participation of training and competition in jumping sports and may be a natural process of change in PT mechanical properties in elite athletes. There were also a number of other confounding variables that may have influenced PT mechanical properties over the course of a season of training and competition other than PT loading history. These primarily relate to the prevention and management of patellar tendinopathy and include the use of the following strategies: pharmacologic agents (Maffulli, Longo, Loppini, & Denaro, 2010) such as non-steroidal anti-inflammatory medication (Fallon, Purdam, Cook, & Lovell, 2008), doxycycline (Fallon, et al., 2008) and cortisone (Kongsgaard, et al., 2009); electrophysiological agents including low intensity pulsed ultrasound (Fu, Shum, Hung, Wong, Qin, & Chan, 2008; Lu, Zheng, Huang, & Qin, 2009; Warden, Metcalf, Kiss, Cook, Purdam, Bennell, & Crossley, 2008) and exercise programs (Frohm, et al., 2007a; Kongsgaard, et al., 2009; Kongsgaard, et al., 2010; Purdam, et al., 2004; Young, et al., 2005). The use of multiple modalities of treatment on PT mechanical properties has not been investigated.

8.3 Future Directions

8.3.1 Lower Limb Motor Control in Elite Athletes in Hopping

An evaluation of kinetics and kinematics in the coronal and transverse planes is an area for future research into lower limb loading during hopping. Although the primary plane of joint motion during hopping is in the anterior-posterior plane, there is also motion in the coronal and transverse planes at each joint in the lower limb that contributes to control and may contribute to the pathogenesis of overuse injury (Kernozek, Torry, Van Hoof, & Tanner, 2005; Yeow, Lee, & Goh, 2009). Mechanical behaviour in coronal and transverse planes has previously been investigated during take-off and landing in volleyball players at the ankle and knee joints (D. P. Richards, et al., 2002; D. P. Richards, et al., 1996) and during a stop-jump task in participants with and without patellar tendinopathy (Edwards, et al., 2010b). However, the mechanical behaviour during hopping, that utilises the SSC in a repetitive manner, has not been investigated. This is of particular interest in single-leg loading strategies that may be hypothesised to be different between sides and groups of athletes with different lower limb loading histories. Although the current study did not demonstrate laterality in the derived variables assessed during the sub-maximal hopping efforts, there may be inter-side differences in hopping tasks that are of greater effort.

Future research into hopping must aim to identify the factors that influence the variability in motor control parameters. Variability may represent the variance of a single motor control strategy or a shift between motor control strategies. For example, knee mechanical work done increased at the knee joint and decreased at the ankle joint when comparing DL hopping on a decline to DL hopping on a flat surface while there was a conservation of vertical leg stiffness in both conditions when hopping frequency was fixed at 1.8Hz. However, the question of whether the observed changes in joint mechanical behaviour and muscle activation characteristics represents variability in a single motor control strategy or a change between control strategies needs to be investigated. Design of testing protocols in future studies should aim to identify factors that influence the variability and shift between motor control strategies. This will provide a greater understanding of how the motor system is regulated and allow for improved assessment and rehabilitation techniques.

8.3.2 Patellar Tendon Mechanical Properties in Elite Athletes

Technological advancements including B-mode ultrasound and intra-tendinous optic fibre insertion have allowed for the measurement of PT mechanical properties *in vivo* and have progressed our understanding of PT mechanics. Further advancement in technology will hopefully allow for a measure of PT mechanics during fast dynamic tasks with greater accuracy that is specific to individuals, non-invasive and inexpensive. Tendon deformation using B-mode ultrasound of the PT during a relatively slow single-leg decline squat (Kongsgaard, et al., 2006) and of the Achilles tendon during hopping (Lichtwark, et al., 2005b; McGuigan, Farris, Wood, Abbas, & Trewartha, 2008) has been used previously.

Direct measure of PT deformation using B-mode ultrasound, during relatively fast dynamic movements such as hopping, is currently difficult due to the relatively low sampling frequency (30 frames/second) of commonly available ultrasound systems and an inability to maintain stability of the ultrasound transducer to the PT as the knee flexes and extends. In the current study no difference in mechanical properties was detected between tendons with and without patellar tendinopathy. However, there may be intratendinous differences in mechanical properties specifically at the site of pathologic tissue. Future studies may focus on developing accurate methods to measure the differential mechanical properties between PT segment lengths in series.

Measuring PT force during dynamics activities (Edwards, et al., 2010b; Zwerver, et al., 2007) was problematic as PT moment arm lengths are derived from estimate techniques based on cadaveric studies and reduce the accuracy of PT force estimates as they are not subject-specific. Intra-tendinous optic fibre insertion methods offer the ability to measure PT force *in vivo* during fast dynamic activities (Dillon, et al., 2008; Elvin, et al., 2009; Finni, et al., 2001); however, this method was invasive and can be painful for the participant. Importantly, the significant findings in this thesis are for the derived variables of PT stiffness and elastic modulus rather than for PT force. Hence, the measure of both PT length and force are necessary to calculate the slope of force-length and stress-strain relationships. Thus, future advancement in imaging and motion analysis technology are necessary to allow calculation of PT mechanics during fast movements. This thesis established an association between PT loading history and mechanical properties. Therefore, future studies must aim to quantify the loading profile of the PT during training, competition and ADL.

8.4 Conclusions

This thesis determined that the magnitude of loading may change at the knee joint during the SSC in elite level jumping athletes. Patellar tendon mechanical properties are similar between elite level athletes that participate in jumping and non-jumping sports and may adapt in relation to the loading history. The strategy for adaptation of joint mechanical behaviour and the PT is specific to the conditions under which load is determined.

APPENDICES

Appendix A - La Trobe University Human Research Ethics Committee Approval

La Trobe University
Faculty of Health Sciences
MEMORANDUM

TO: Dr Jill Cook School of Physiotherapy

SUBJECT: *Reference:* **FHEC06/023**

Student or Other Investigator: Ms Amitabh Gupta

Title: **Sensory-motor interaction of the lower limb during impact loading: Tendon mechanical properties**

DATE: 13 April, 2006

The Faculty Human Ethics Committee (FHEC) has considered the above project. You may now proceed.

Please note that Informed Consent forms need to be retained for a minimum of 7 years. Please ensure that each participant retains a copy of the Informed Consent form. Researchers are also required to retain a copy of all Informed Consent Forms separately from the data. The data must be retained for a period of 15 years.

Please note that any modification to the project must be submitted in writing to FHEC for approval. You are required to provide an annual report (where applicable) and/or a final report on completion of the project. A copy of the progress/final report is enclosed. Would you please return the completed form to Ms Natalie Humphries, Secretary, FHEC, Faculty of Health Sciences Office, La Trobe University, Victoria 3086.

A copy of this memorandum is enclosed for you to forward to the student(s) concerned.

Natalie Humphries

Natalie Humphries
Secretary
Faculty Human Ethics Committee
Faculty of Health Sciences

Appendix B - Australian Institute of Sport Human Research Ethics Committee Approval

Australian Institute of Sport

MINUTE

TO: Amitabh Gupta **CC:** Craig Purdam

FROM: Helene Kay

SUBJECT: Approval from AIS Ethics Committee **DATE:** 22nd February 2006

I am pleased to inform you that at the last meeting of the AIS Ethics Committee held on the 21st February 2006, the Committee saw no ethical reason why your project ***"Sensory-motor interaction of the lower limb during impact loading – Tendon mechanical properties in athletes"*** should not proceed.

The approval number for this project is 20060203.

It is a requirement of the AIS Ethics Committee that the Principal Researcher (you) advise all researchers involved in the study of Ethics Committee approval and any conditions of that approval. You are also required to advise the Ethics Committee immediately (via the Secretary) of:

any proposed changes to the research design,
any adverse events that may occur,

Researchers are required to submit **annual status reports** to the secretary of the AIS Ethics Committee until completion of the project. Details of status report requirements are contained in the "Guidelines" for ethics submissions.

Failure to comply with the above will render ethics approval null and void.

If you have any questions regarding this matter, please don't hesitate to contact me on (02) 6214 1816.

Sincerely
Helene Kay
On behalf of

John Williams
Secretary, AIS-EC

Appendix C – Informed Consent Form

Project Title: Sensory-motor interaction of the lower limb during impact loading – tendon mechanical properties in athletes.

Principal researcher: Amitabh Gupta

This is to certify that I, ________________________________hereby agree to (give permission to have my child) participate as a volunteer in a scientific investigation as an authorised part of the research program of the Australian Sports Commission under the supervision of _________________________________.

The investigation and my (child's) part in this investigation have been defined and fully explained to me by Amitabh Gupta and I understand the explanation. A copy of the procedures of this investigation and a description of any risks and discomforts has been provided to me and has been discussed in detail with me.

I have been given an opportunity to ask whatever questions I may have had and all such questions and inquiries have been answered to my satisfaction.
I understand I am free to deny any answers to specific items or questions in interviews or questionnaires.
I understand that I am (the child) is free to withdraw consent and to discontinue participation in the project or activity at any time.
I understand that any data or answers to questions will remain confidential with regard to my (child's) identity.
I certify to the best of my knowledge and belief, I have (the child has) no physical or mental illness or weakness that would increase the risk to me (him/her) of participating in this investigation.
I am (the child is) participating in this project of my (his/her) own free will and I have (the child has) not been coerced in any way to participate.

Signature of subject: ______________________________________ Date: ___/___/___

Signature of parent or
Guardian of minor (under 16 years) __________________________ Date: ___/___/___

I, the undersigned was present when the study was explained to the subject in detail and to the best of my knowledge and belief it was understood.

Signature of researcher : ___________________________________ Date: ___/___/___

Appendix D - Equipment and instrumentation used in chapter 2

Electromyography(EMG): Myosystem 1400A, 16-channel fixed wire system, Noraxon Inc., Az, USA.

Power requirements: 90-240v AC, 50/60 Hz.

Hardware filters: No notch filter

· All surface EMG (sEMG) leads have a 1st order high-pass filter set to 10 Hz ±10% cutoff.

· All sEMG channels have 8^{th} order Butterworth low-pass filters set to either 500 Hz of 1,000 Hz ±2% cutoff.

· Baseline noise <1μV RMS

· Input impedance > 100MOhm

· Common Mode Rejection Ratio (CMRR) > 100 dB

· Range 0 to ±7 mV

· Isolation > 3,000 Volts

Analogue outputs:

· 25 pin D-sub connector

· All EMG channels gain of 500 (fixed)

· All outputs ±5 Volts max

Data Acquisition System

· 12-bit resolution

· Noise <2 LSB

· 20 channels (16 sEMG & 4 auxiliary)

· Individual gains settings for each channel ((500) x1, x2, x4, x5, x8, x10)

Physical dimensions:

· 28.0 x 19.7 x 10.2 cm; 1.8 kg

Preamplified Leads:

· Differential gain 0f 500 ±2%

· Lead length of 121.92 cm

· Lead connector 4 pin Alpha Products 300/231C1090040

· Mating connector 4 pin Alpha Products 300/231C1090240

· Connector pinout including a preamp output (white), +5volts (red), -5 volts (black) and common (green)

Appendix D (continued) - Equipment and instrumentation used in chapter 2

Motion Analysis: VICON Motion Systems, Ox, UK

- High resolution cameras (MX40) (4 megapixels): Sensor resolution: 2352 h x 1728 v
- Low resolution cameras (MX 13) (1.3 megapixels): Sensor resolution: 1280 h x 1024 v

Kinetics: Two Multicomponent Force Plates, 900 x 600 x 100 mm, 9287BA, Kistler Instrument Corp., NY, USA.

Built in 8 channel charge amplifier (tyoe 9287CA)

- Calibrated range: F_x, F_y 0 to 5 kN; F_z 0 to 20 kN
- Calibrated partial range: F_x, F_y 0 to 1.25 kN; F_z 0 to 5 kN
- Sensitivity range 1: F_x, F_y ≈40 mV/N(±0.5%); F_y ≈18 mV/N(±0.5%)
- Sensitivity range 4: F_x, F_y ≈2 mV/N(±0.5%); F_y ≈0.9 mV/N(±0.5%)
- Ratio ranges 1:2:3:4 – 1:5:10:20
- Threshold: <250mN
- Drift < ±10 mN/s
- Supply voltage 10 to 30 VDC
- Supply current ≈45 mA
- Output voltage 0 ±5 V
- Output current -2 to 2 mA
- Control inputs (optocoupler) 5 to 45 V; 0.4 to 4.4 mA

Appendix E - Marker placements for the VICON plug-in-gait model in chapter 2

Anatomical Landmark	No. of markers	Sides
Front of head	2	left & right
Back of head	2	left & right
C7 spinous process	1	
T10 spinous process	1	
Jugular notch	1	
Sternum	1	
Scapula	1	right
Acromio-clavicular joint	2	left & right
Lateral epicondyle of humerus	2	left & right
Dorsum of 2nd metacarpal joint	2	left & right
Radial styloid process	2	left & right
Ulna styloid process	2	left & right
Posterior superior iliac spine	2	left & right
Anterior superior iliac spine	2	left & right
Thigh marker (2/3 distance between greater trochanter and lateral femoral epicondyle)	2	left & right
Lateral femoral epicondyle	2	left & right
Shank marker (2/3 the distance from the lateral femoral epicondyle to the lateral malleolus)	2	left & right
Lateral malleolus	2	left & right
Dorsum of the 2nd metatarsophalangeal joint	2	left & right
Posterior aspect of the calcaneum (maintained at the same transverse plane to the previous marker)	2	left & right

Appendix F – Inverse Solution for calculation of joint moments in chapter 2

The joint moment is the net summation of all muscle activity at each joint. A link segment model was developed to determine joint moments and was based on the following assumptions (Winter, 1990):

1. Each segment has a fixed mass located as a point mass as its centre (COM).
2. The location of each segment's COM remains fixed during movement.
3. The joints are considered to be hinge joints.
4. The mass moment of inertia of each segment about its mass centre (or about its proximal or distal joints) is constant during movement.
5. The length of each segment remains constant during the movement.

For the inverse solution of the lower limb a three segment model was used representing the foot, shank and thigh with a mass and moment of inertia at each segment's COM (Figure F.1).

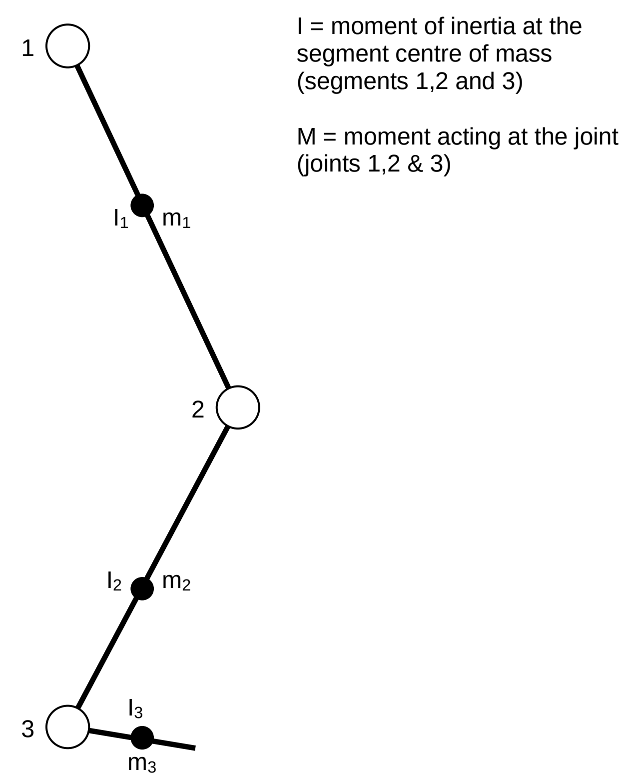
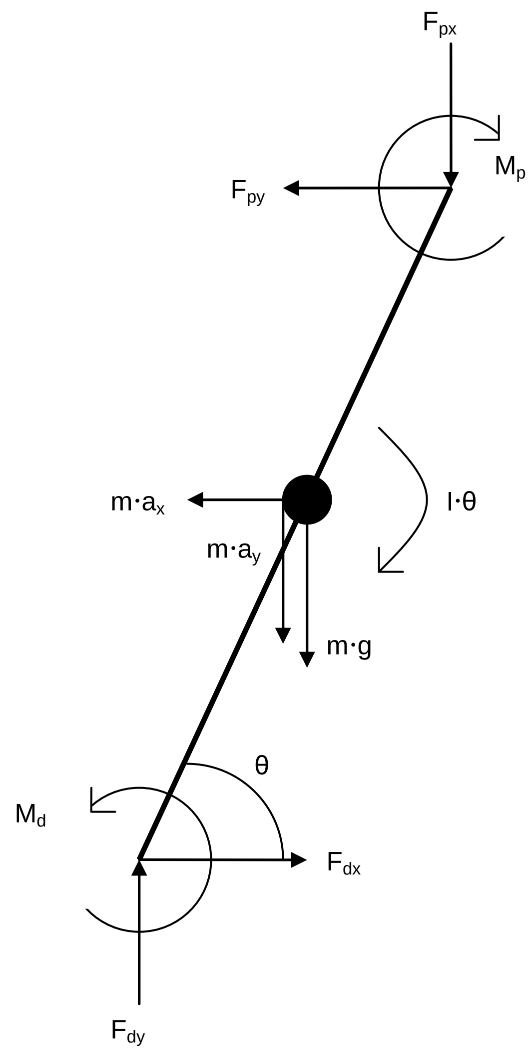


Figure F.1 The free-body diagram illustrates the reaction forces and moments of force acting at each joint.

Segmental kinematics and reaction force data are combined to allow calculation of moments and reaction forces. Thus, when the reaction force at the distal end, net muscle moment acting at the distal joint and kinematic data (acceleration of segment centre of mass (COM), angle of segment in plane of movement, angular acceleration of segment in plane of movement) are known, the net reaction force at the proximal joint may be calculated (Figure F.2).



Nomenclature:
m = mass of the segment

I = moment of inertia of the segment about the centre of mass

θ = angular acceleration of the segment in the plane of movement

a_x, a_y = acceleration of segment centre of mass

g = gravity

F_{pz}, F_{py} = reaction forces acting at proximal end

F_{dx}, F_{dy} = reaction of forces acting at distal end

M_p = net moment acting at the proximal joint

M_d = net moment acting at the distal joint

Figure F.2 Free body diagram of a single segment showing reaction force and gravitational forces, net moments of force and all linear and angular accelerations.

To calculate reaction forces in the proximal joint and the net muscle moment acting at a proximal joint, the following equations are used:

1. $\Sigma F_x = ma_x$

$F_{px} – F_{dx} = ma_x$

solve for F_{px}

2. $\Sigma F_y = ma_y$

$F_{py} – F_{dy} – mg = ma_y$

solve for F_{py}

3. $\Sigma M = I.\theta$

$M_p – F_{dy}\,x – F_{py}\,x = 0$

Where, I is the moment of inertia about the COM, θ is the angular acceleration of the segment in the plane of movement, F_{dy} x is the length from COM to application of F_{dy} and F_{py}, x is the length from COM to proximal joint.

Solve for M_p

Thus, the measure of force in the *x* and *y* axes via the use of a force plate allows for calculation of joint moment at the ankle, knee and hip via an inverse solution by solving for the joint moment at the proximal joint for each segment, commencing with the foot.

Appendix G - Calculation of Joint Moments when hopping on a decline surface (chapter 2)

G.1 Introduction

The centre of pressure (COP) was more accurately represented by placing the force plate directly under the foot in the motion analysis as during flat surface hopping (Figure G.1.1). This allows for a more accurate calculation of joint moments during decline hopping.

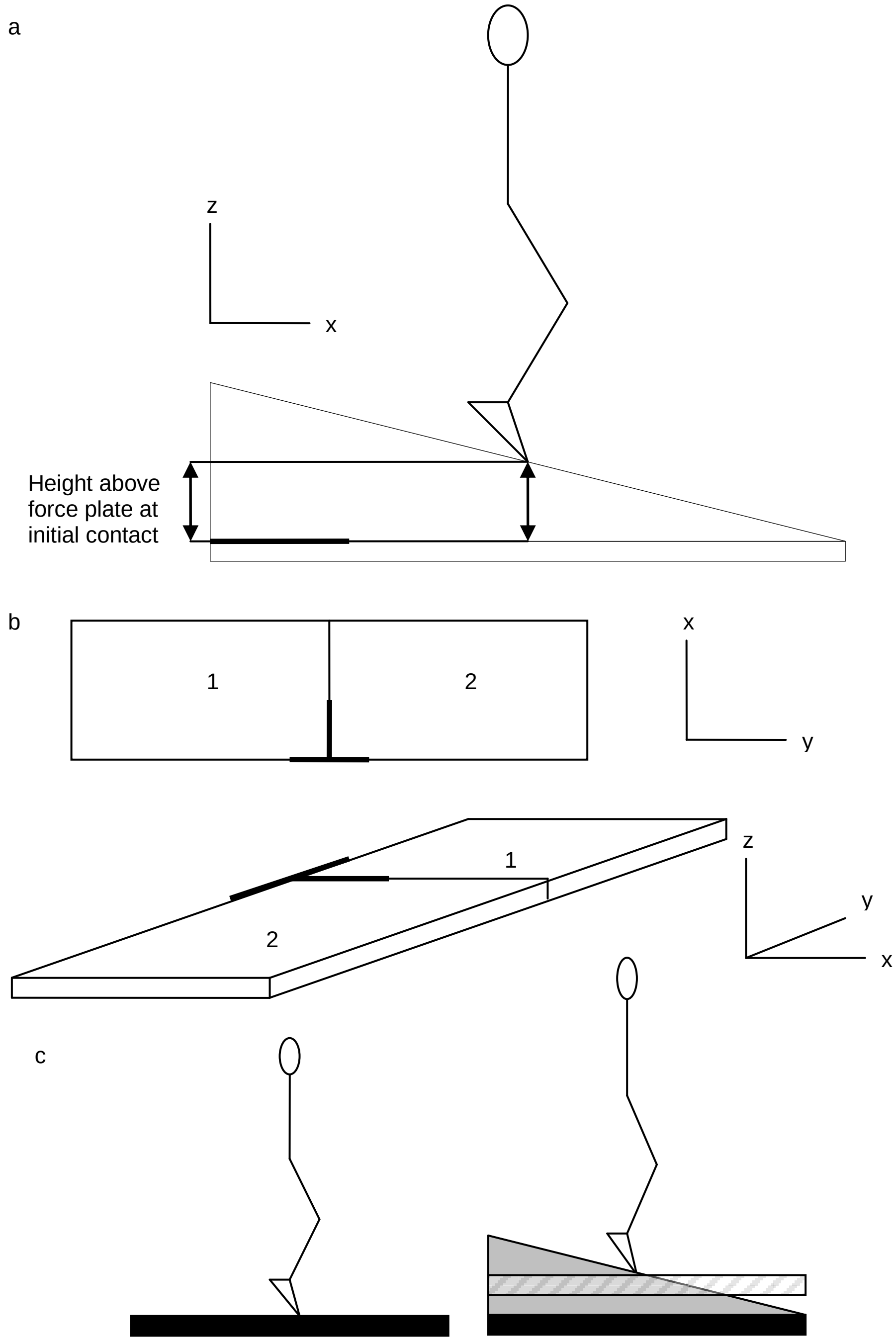


Figure G.1.1 Schematic representation of a participant hopping on a decline surface where they are observed to contact the decline surface above the force plate as seen along the y axis (a). The 'T frame' (thick lines) was placed with the vertical section of the frame along the x axis and the cross bar of the frame along the y axis (b). A participant hopping on a flat surface (c - image left) and on a decline surface (c - image right) where the force plate has been raised to a position directly under the foot (shaded rectangle).

During hopping the first point of contact was the ball of the foot. The toe-marker placed on the dorsal surface of the 2nd metatarsophalangeal joint, represents the only part of the plug-in-gait model that approximates the point of contact of the body to the ground. Thus, the height (z axis) between the toe marker and the global origin was estimated as being the same during static standing (Figure G.1.2) and at initial contact during decline hopping (Figure G.1.3) for each side.

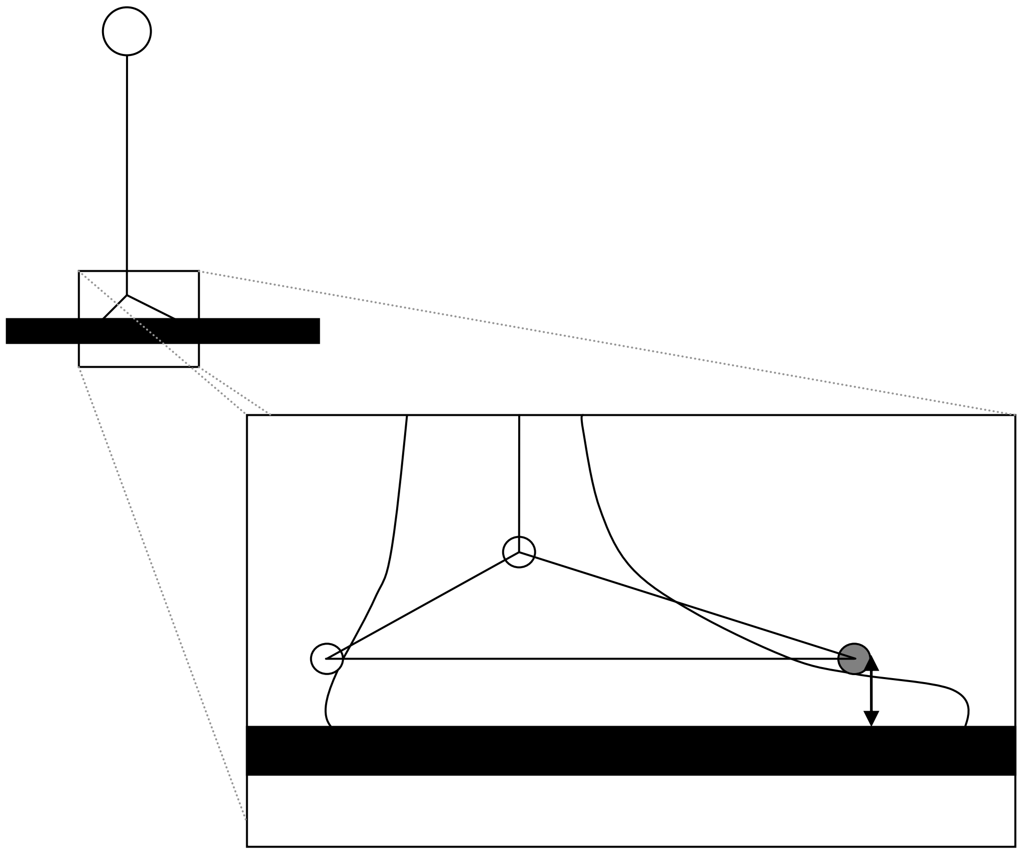


Figure G.1.2 Schematic representation of the foot markers in the plug-in-gait model. The distance in the z axis (double ended arrow) from the global origin to the toe-marker (shaded grey circle) is recorded from the static calibration trial for each participant for each side.

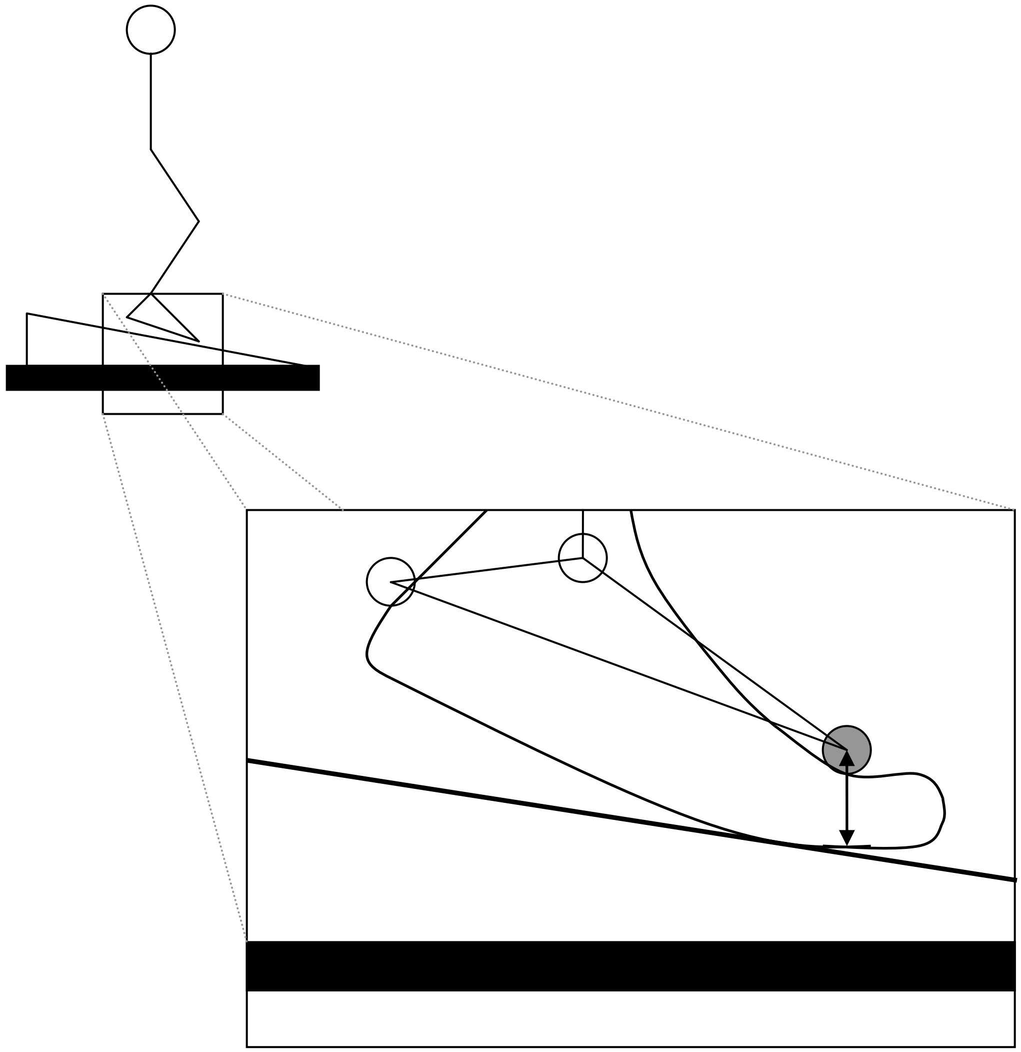


Figure G.1.3 Schematic representation of the foot markers in the plug-in-gait model. The distance in the z axis (double ended arrow) from the global origin to the toe-marker recorded from the static calibration is maintained as the same distance during initial contact for decline hopping on each side.

To adjust for an accurate measure of COP a pilot study was performed to examine the concordance of estimates of joint moment using two methods. Firstly, correction of joint moment calculation was performed by correcting for the height difference (z axis) for the toe marker above the force plate on each side (left and right) for each of the 10 successive hops selected for analysis. Secondly, correction of joint moment calculation was performed by correcting for the height difference (z axis) for the toe marker above the force plate on each side (left and right) as a mean of the height correction for the 10 successive hops selected for analysis. Thus, the pilot study focuses on comparing joint moments as a result for correcting force plate position for each of the 10 hops or using the mean value across the 10 hops.

G.2 Method

To be able to place the force plate on the left and right sides directly under the left and right toe-markers respectively, it was necessary to calculate the corrected height ($z_{corrected}$) by deducting the static height (z_{static}) from the original height of the toe marker ($z_{original}$) during decline surface hopping at initial contact. The point of initial contact for each hop was identified using the same method as described in chapter 3 (section 3.3.6).

Once the corrected height of the left and right toe-markers was calculated, the left and right force plates were adjusted to the new height in the Nexus software (version 1.4). Ten consecutive hops were chosen for each side during each trial on a decline surface for each participant. An adjustment to the position of the force plates in the z axis for each hop on the left and right sides most closely corrects for the COP measure and therefore provides the most accurate measure of joint moments on the left and right sides for each individual hop. Thus, this correction to the height of the force plate was made for each of the ten hops on the left and right sides for each trial. Then, following the same procedure as outlined in chapter 3 (section 3.3), the makers were labeled, gaps were filled and data filtered. Once filtered, the dynamic model was run and files containing joint moments were exported for data analysis. This data processing procedure of correcting the force plate height on left and right sides for each of the ten hops was performed for decline surface hopping trials for 3 of the 9 participants that were included in the study on motor control during hopping (chapter 2).

The mean height correction over the ten hops was also calculated for left and right sides of each trial. This value was also used to correct the height of the force plate followed by labeling of the model, gap filling, data filtering and the dynamic model being run. Data on joint moment was then exported to a spreadsheet (Microsoft Excel 2003) for analysis.

G.2.1 Data Analysis

Joint moments were derived using two methods. Firstly, the reference method (M_R) where each hop on each side had the force plate height adjusted on left and right sides. Secondly, the mean height adjustment (M_M) of the force plates was calculated and used as a value for height adjustment on left and right force plates.

The correlation coefficient (Pearson's R) was calculated comparing joint moments using M_R and M_M. Linear regression analysis was performed to compare joint moments calculated from the two methods, M_R and M_M, for left and right force plates. All data points were entered into a spreadsheet (W.G. Hopkins, 2000). When significant linear correlations were identified, the mean and 95% confidence interval (CI) of the regression coefficient (slope) and Y intercept for elongation were determined. Typical error (SDdiff/ $\sqrt{2}$, where SDdiff is the standard deviation of the difference scores) and 95% CI were determined for the paired comparisons of M_R against M_M (W.G. Hopkins, 2000).

G.3 Results

The ankle, knee and hip joint moments were calculated for left and right sides, over four trials for three participants providing a possible of 72 data points. Five data points were missing providing a total of 67 data points for comparison of M_R and $M_{M.}$

When comparing joint moments using M_R to M_M, Pearson's R = 0.95. For linear regression of M_R against M_M the regression coefficient (slope) for moment was (mean (95% CI)) 0.975 (0.894 to 1.056) and the intercept of the line of regression was 0.018 $kN.mm.kg^{-1}$ (-0.138 to 0.175) for moment. This demonstrates a strong linear relationship between M_R and M_M with a regression coefficient approaching 1.0 and an intercept approaching zero (Figure G.3.1). Typical error was mean 0.307$kN.mm.kg^{-1}$ (95%CI) (0.262 to 0.37).

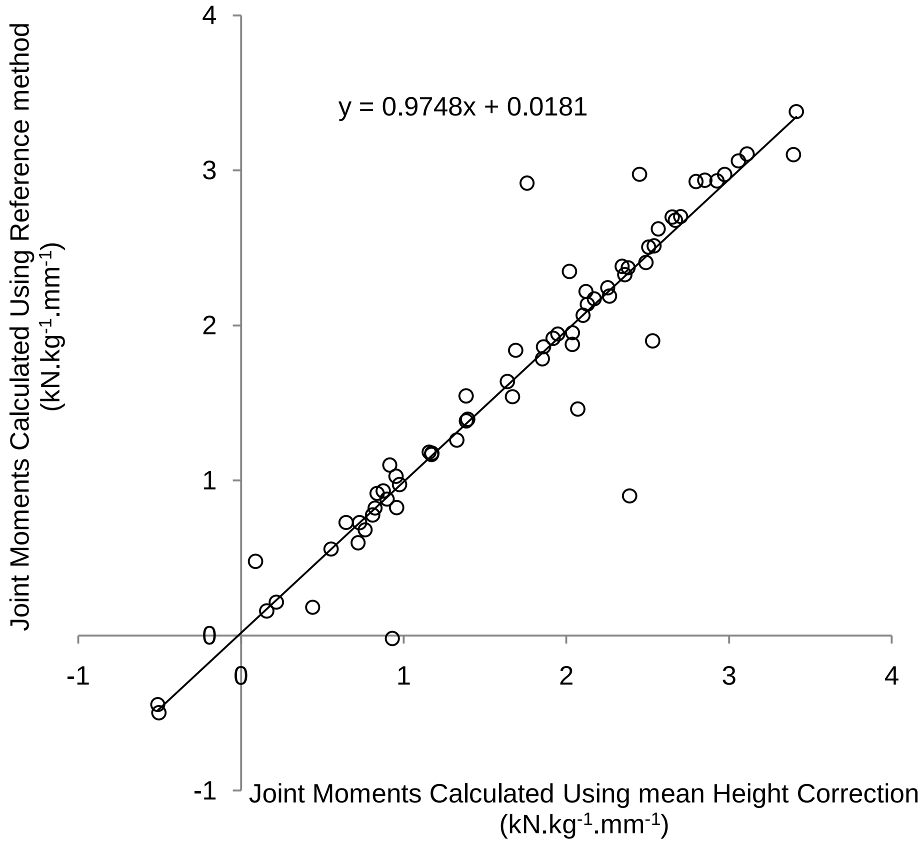


Figure G.3.1 Linear regression comparing the resultant joint moments calculated from the reference method (M_R) and mean height correction for force plates (M_M). There is a strong linear relationship with a regression coefficient of 0.975 and an intercept of 0.018.

G.4 Discussion

Using the mean height correction over 10 hops yielded joint moments that are strongly correlated to joint moments using a reference method that corrects the height of the force plate for each hop separately. Thus, calculating the mean height correction of the left and right force plates to yield joint moments for all participants in a study including the task of hopping on a decline surface was a valid and appropriate method for correction of joint moments. This pilot study demonstrated a strong linear relationship between two methods that did not differ from the line of identity (i.e.; slope = 0.97 and intercept = 0.02)

The method of adjusting the height of the force plate, performed on the data for 3 participants and determined by the position of the toe-market relative to the floor surface was deemed representative of the height of the contact point of the participant on the decline board to the force plate surface. It was concluded that the method to adjust for the height of the participant hopping on a decline surface was valid when derived from the average height adjustment of all hops included in each trial and that this was an accurate estimate for correction for height of the decline board that to reduce data processing time.

Appendix H - Statistical results for inter-side comparisons in chapter 3

The following results are for the side to side comparison for each of the main derived variables. Note that 6 of the multiple comparisons were below 0.05 and none of the comparisons were below 0.01. In summary, there was a clear indication that this study did not display any systematic difference between the left and right sides for the sample population for any of the derived variables.

Table H.1 Paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for inter-side comparison of hopping frequency across all hopping conditions.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL Flat S	-0.0007 (0.0053)	0.0018 [-0.005 to 0.004]	8	-0.368	0.722	0.01
DL Dec S	-0.0009 (0.004)	0.0014 [0.004 to 0.0022]	8	-0.659	0.528	0.01
AL Flat S	0.000 (0.0088)	0.0029 [-0.0068 to 0.0068]	8	0.000	1.000	0.01
AL Dec S	0.0001 (0.0044)	0.0015 [-0.0033 to 0.0035]	8	0.76	0.941	0.01
DL Flat F	-0.0002 (0.0015)	0.0005 [-0.0014 to 0.0009]	8	-0.45	0.665	0.01
DL Dec F	0.0004 (0.0026)	0.0009 [-0.0015 to 0.0024]	8	0.522	0.616	0.02
AL Flat F	0.0056 (0.0161)	0.0054 [-0.0068 to 0.0179]	8	1.034	0.331	0.04
AL Dec F	-0.0058 (0.011)	0.0037 [-0.0048 to 0.0035]	8	-1.578	0.153	0.10

DL – Double-Leg; AL – Alternate-Leg; S – Self-Selected Pace; F – Forced Pace; Dec – Decline

Table H.2 Paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for inter-side comparison of vertical height displacement (mm) across all hopping conditions.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95% CI]	df	t	p	1-β
DL Flat S	0.99 (2.83)	0.94 [-1.19 to 3.16]	8	1.046	0.326	0.04
DL Dec S	-0.62 (2.86)	0.62 [2.05 to 8.00]	8	-1.008	0.343	0.02
AL Flat S	0.71 (5.35)	1.78 [-3.40 to 4.82]	8	0.398	0.701	0.01
AL Dec S	4.64 (4.06)	1.54 [-8.39 to 0.88]	6	-3.021	0.023	0.56
DL Flat F	0.24 (1.45)	0.48 [-0.87 to 1.36]	8	0.503	0.629	0.02
DL Dec F	-0.59 (1.19)	0.40 [-1.50 to 0.32]	8	-1.490	0.175	0.09
AL Flat F	0.74 (9.57)	3.19 [-6.62 to 8.09]	8	0.231	0.823	0.01
AL Dec F	-1.89 (12.23)	4.08 [-11.29 to 7.51]	8	0.463	0.656	0.02

DL – Double-Leg; AL – Alternate-Leg; S – Self-Selected Pace; F – Forced Pace; Dec – Decline

Table H.3 Paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for inter-side comparison of peak vertical ground reaction force (N) across all hopping conditions.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL Flat S	0.046 (0.182)	0.061 [-0.094 to 0.185]	8	0.756	0.471	0.03
DL Dec S	-0.106 (0.191)	0.064 [0.252 to 0.041]	8	-1.663	0.135	0.11
AL Flat S	0.005 (0.051)	0.017 [-0.034 to 0.044]	8	0.294	0.776	0.01
AL Dec S	0.008 (0.144)	0.048 [-0.103 to 0.118]	8	0.164	0.874	0.01
DL Flat F	0.017 (0.118)	0.039 [-0.074 to 0.108]	8	0.434	0.675	0.01
DL Dec F	-0.024 (0.129)	0.043 [-0.123 to 0.075]	8	-0.552	0.596	0.02
AL Flat F	-0.015 (0.067)	0.022 [-0.067 to 0.037]	8	-0.658	0.529	0.02
AL Dec F	-0.090 (0.138)	0.046 [-0.196 to 0.015]	8	-1.97	0.084	0.16

DL – Double-Leg; AL – Alternate-Leg; S – Self-Selected Pace; F – Forced Pace; Dec – Decline

Table H.4 Paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for inter-side comparison of joint excursion (ankle, knee and hip) (degrees) during braking phase for each hopping condition.

Hopping condition		Mean Difference (SD)	Standard error of the mean [95% CI]	df	t	p	1-β
DL Flat S	Ankle	1.56 (4.52)	1.51 [-1.94 to 5.01]	8	1.019	0.338	0.04
	Knee	-1.2 (2.83)	0.94 [-3.38 to 0.97]	8	-1.276	0.238	0.06
	Hip	-0.38 (1.74)	0.58 [-1.72 to 0.96]	8	-0.654	0.531	0.02
DL Dec S	Ankle	5.47 (9.70)	3.23 [-1.99 to 12.93]	8	1.692	0.129	0.11
	Knee	-0.63 (1.78)	0.59 [-2.00 to 0.74]	8	-1.064	0.318	0.04
	Hip	-2.86 (7.64)	2.54 [-8.73 to 3.02]	8	-1.121	0.295	0.04
AL Flat S	Ankle	3.87 (7.83)	2.61 [-2.15 to 9.89]	8	1.483	0.176	0.08
	Knee	3.59 (4.85)	1.62 [-0.13 to 7.32]	8	2.223	0.057	0.21
	Hip	1.33 (1.88)	0.63 [-0.12 to 2.78]	8	2.122	0.067	0.19
AL Dec S	Ankle	1.64 (5.22)	1.74 [-2.37 to 5.65]	8	0.943	0.373	0.04
	Knee	-0.26 (4.44)	1.48 [-3.67 to 3.15]	8	-0.176	0.864	0.01
	Hip	-0.89 (1.98)	0.66 [-2.41 to 0.64]	8	-1.343	0.216	0.07
DL Flat F	Ankle	4.58 (6.56)	2.19 [-0.46 to 9.63]	8	2.094	0.070	0.19
	Knee	-0.37 (3.08)	1.03 [-2.74 to 1.99]	8	-0.362	0.726	0.01
	Hip	0.12 (2.28)	0.76 [-1.63 to 1.87]	8	0.159	0.877	0.01
DL Dec F	Ankle	3.77 (5.32)	1.77 [-0.32 to 7.86]	8	2.123	0.066	0.19
	Knee	-0.83 (2.77)	0.92 [-2.96 to 1.30]	8	-0.899	0.395	0.03
	Hip	-0.44 (1.96)	0.65 [-1.95 to 1.06	8	-0.68	0.515	0.02
AL Flat F	Ankle	5.72 (9.68)	3.23 [-1.72 to 13.17]	8	1.772	0.114	0.13
	Knee	2.20 (4.96)	1.65 [-1.61 to 6.01]	8	1.331	0.220	0.07
	Hip	-0.02 (2.53)	0.84 [-1.96 to 1.93]	8	-0.019	0.985	0.01
AL Dec F	Ankle	4.16 (5.07)	1.69 [-0.26 to 8.06]	8	2.457	0.039*	0.27
	Knee	0.88 (3.23)	1.08 [-1.61 to 3.36]	8	0.812	0.440	0.03
	Hip	-0.70 (2.29)	0.76 [-2.46 to 1.06]	8	-0.922	0.384	0.03

DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace, F – Forced Pace, Dec – Decline; *- p < 0.05

Table H.5 Paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for inter-side comparison of normalised joint moments (ankle, knee and hip) ($Nm.kg^{-1}$) during braking phase for each hopping condition.

Hopping condition		Mean Difference (SD)	Standard error of the mean [95% CI]	df	t	p	1-β
DL	Ankle	0.069 (0.53)	0.18 [-0.34 to 0.48]	8	0.395	0.703	0.01
Flat S	Knee	-0.01 (0.67)	0.23 [-0.54 to 0.52]	8	-0.052	0.960	0.01
	Hip	-0.56 (0.73)	0.24 [-1.12 to 0.01]	8	-2.278	0.052	0.23
DL	Ankle	0.14 (0.45)	0.15 {-0.20 to 0.49]	8	0.957	0.366	0.04
Dec S	Knee	-0.29 (0.72)	0.24 [-0.84 to 0.26]	8	-1.232	0.253	0.06
	Hip	-0.25 (0.49)	0.16 [-0.63 to 0.13]	8	-1.522	0.167	0.09
AL	Ankle	0.22 (0.42)	0.14 -0.10 to 0.55]	8	1.593	0.150	0.10
Flat S	Knee	-0.10 (0.51)	0.17 [-0.49 to 0.29]	8	-0.604	0.563	0.02
	Hip	-0.16 (0.38)	0.13 [-0.45 to 0.134]	8	-1.246	0.248	0.06
AL	Ankle	0.20 (0.37)	0.12 [-0.09 to 0.48]	8	1.587	0.151	0.10
Dec S	Knee	-0.11 (0.81)	0.27 [-0.723 to 0.51]	8	-0.394	0.704	0.01
	Hip	-0.46 (0.59)	0.20 [-0.91 to <-0.01]	8	-2.308	0.05	0.24
DL	Ankle	0.20 (0.39)	0.13 [-0.10 to 0.50]	8	1.527	0.165	0.09
Flat F	Knee	-0.027 (0.63)	0.21 [-0.51 to 0.46]	8	-0.132	0.898	0.01
	Hip	-0.47 (0.43)	0.14 [-0.80 to -0.14]	8	-3.271	0.011*	0.5
DL	Ankle	0.23 (0.39)	0.13 [-0.07 to 0.53]	8	1.786	0.112	0.13
Dec F	Knee	-0.39 (0.89)	0.30 [-107 to 0.30]	8	-1.306	0.228	0.07
	Hip	-0.27 (0.68)	0.23 [-0.80 to 0.25]	8	-1.200	0.265	0.05
AL	Ankle	-0.09 (0.93)	0.31 [-0.80 to 0.63]	8	-0.285	0.783	0.01
Flat F	Knee	-0.073 (0.53)	0.18 [-0.48 to 0.34]	8	-0.409	0.693	0.01
	Hip	-0.61 (0.76)	0.25 [-1.20 to -0.03]	8	-2.415	0.042*	0.26
AL	Ankle	0.13 (0.39)	0.13 [-0.17 to 0.43]	8	0.998	0.347	0.04
Dec F	Knee	-0.26 (0. 09)	0.30 [-0.95 to 0.43]	8	-0.872	0.409	1.0
	Hip	-0.38 (0.40)	0.133 [-0.68 to -0.07]	8	-2.847	0.022*	0.38

DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace, F – Forced Pace, Dec – Decline; *- $p < 0.05$

Table H.6 Paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for inter-side comparison of normalised joint stiffness ($Nm.kg^{-1}.rad^{-1}$) (ankle, knee and hip) during braking phase for each hopping condition.

		Mean Difference (SD)	Standard error of the mean [95% CI]	df	t	p	1-β
DL	Ankle	-0.21 (0.82)	0.27 [-0.83 to 0.42]	8	-0.766	0.466	0.03
Flat S	Knee	0.21 (1.59)	0.53 [-1.01 to 1.44]	8	0.401	0.699	0.01
	Hip	-16.84 (34.50)	11.50 [-4335.57 to 9.68]	8	-1.464	0.181	0.08
DL	Ankle	-2.6 (8.78)	2.93 [-9.61 to 3.89]	8	-0.977	0.357	0.03
Dec S	Knee	-0.67 (1.53)	0.51 [-1.84 to 0.51]	8	-1.305	0.228	0.07
	Hip	-1.05 (2.65)	0.89 [-3.09 to 0.98]	8	-1.194	0.267	0.05
AL	Ankle	-0.82 (2.42)	0.81 [-2.68 to 1.03]	8	-1.023	0.336	0.04
Flat S	Knee	-10.70 (19.73)	65.77 [-25.86 to 4.48]	8	-1.625	0.143	0.1
	Hip	-20.89 (70.68)	23.56 [-75.13 to 33.44]	8	-0.887	0.401	0.03
AL	Ankle	-0.16 (1.24)	0.41 [-1.11 to 0.80]	8	-0.381	0.713	0.01
Dec S	Knee	-0.20 (5.70)	1.90 [-4.58 to 4.17]	8	-0.107	0.917	0.01
	Hip	0.04 (0.24)	0.08 [-0.14 to 0.02]	8	0.571	0.583	0.02
DL	Ankle	-0.23 (0.79)	0.26 [-0.84 to 0.38]	8	-0.873	0.408	0.03
Flat F	Knee	-0.10 (1.72)	0.57 [-1.43 to 1.22]	8	-0.179	0.862	0.01
	Hip	-4.80 (5.05)	1.68 [-8.68 to -0.92]	8	-2.855	0.021*	0.38
DL	Ankle	-0.02 (0.62)	0.21 [-0.50 to 0.45]	8	-0.114	0.912	0.01
Dec F	Knee	-0.73 (1.70)	0.57 [-2.03 to 0.58]	8	-1.288	0.234	0.06
	Hip	-1.08 (3.19)	1.06 [-3.53 to 1.38]	8	-1.012	0.341	0.04
AL	Ankle	-0.91 (1.81)	0.60 [-2.30 to 0.49]	8	-1.499	0.172	0.09
Flat F	Knee	-0.73 (1.58)	0.53 [-1.95 to 0.48]	8	-1.397	0.200	0.07
	Hip	-13.63 (28.80)	9.60 [-35.77 to 8.50]	8	-1.420	0.193	0.08
AL	Ankle	-0.27 (0.64)	0.21 [-0.77 to 0.23]	8	-1.258	0.244	0.06
Dec F	Knee	-0.84 (2.03)	0.68 [-2.40 to 0.72]	8	-1.246	0.248	0.06
	Hip	-1.77 (2.94)	0.98 [-4.03 to 0.50]	8	-1.801	0.109	0.13

DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace, F – Forced Pace, Dec – Decline; *- $p < 0.05$

Table H.7 Paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for inter-side comparison of peak activation amplitude difference scores (%) for tibialis anterior, soleus, medial gastrocnemius, biceps femoris, vastus lateralis and gluteus medius.

Muscle		Mean Difference (SD)	Standard error of the mean [95% CI]	df	t	p	1-β
DL Dec S	TA	26.77 (67.66)	22.55 [-25.24 to 78.78]	8	1.187	0.269	0.05
	Sol	-20.67 (62.15)	20.72 [-68.44 to 27.10]	8	-0.998	0.348	0.04
	MG	2.92 (36.05)	12.02 [-24.79 to 30.63]	8	0.243	0.814	0.01
	BF	-2.97 (29.89)	10.57 [-27.96 to 22.01]	7	-0.281	0.786	0.01
	VL	22.42 (52.20)	17.40 [-17.70 to 62.54]	8	1.289	0.234	0.06
	GM	-32.15 (112.17)	37.39 [-118.38 to 54.07]	8	-0.86	0.415	0.03
AL Flat S	TA	11.36 (46.55)	14.38 [-21.81 to 44.51]	8	0.789	0.915	0.02
	Sol	-13.46 (21.11)	7.27 [-30.23 to 3.31]	8	-1.851	0.101	0.15
	MG	-10.12 (24.93)	8.31 [-29.28 to 9.05]	8	-1.217	0.258	0.06
	BF	21.17 (77.19)	27.29 [-43.36 to 85.70]	7	0.776	0.463	0.03
	VL	1.59 (27.74)	9.25 [-19.73 to 22.92]	8	0.172	0.867	0.01
	GM	24.10 (65.85)	21.95 [-26.52 to 74.72]	8	1.098	0.304	0.05
AL Dec S	TA	33.76 (58.69)	19.56 [-11.35 to 78.87]	8	1.726	0.706	0.12
	Sol	-4.67 (35.69)	11.90 [-32.10 to 22.76]	8	-0.393	0.705	0.01
	MG	-7.03 (17.14)	5.71 [-20.20 to 6.15]	8	-1.23	0.254	0.06
	BF	24.90 (58.85)	20.81 [-24.29 to 74.10]	7	1.197	0.27	0.06
	VL	7.52 (18.23)	6.08 [-6.49 to 21.54]	8	1.238	0.251	0.06
	GM	-24.78 (181.17)	60.39 [-164.04 to 114.08]	8	-0.41	0.692	0.01
DL Flat F	TA	38.31 (122.55)	40.85 [-55.89 to 132.51]	8	0.938	0.866	0.04
	Sol	67.83 (246.29)	82.10 [-121.49 to 257.15]	8	0.826	0.433	0.03
	MG	3.60 (27.20)	9.07 [-17.31 to 24.51]	8	0.397	0.702	0.01
	BF	50.31 (138.79)	49.07 [-65.72 to 166.34]	7	1.025	0.339	0.05
	VL	0.22 (31.41)	10.47 [-23.93 to 24.36]	8	0.021	0.984	0.01
	GM	2.42 (32.95)	10.98 [-22.92 to 27.75]	8	0.22	0.831	0.01
DL Dec F	TA	43.39 (107.18)	35.73 [-39.00 to 125.78]	8	-0.75	0.214	0.06
	Sol	60.65 (188.01)	62.67 [-83.87 to 205.16]	8	0.968	0.362	0.04
	MG	6.55 (26.73)	8.91 [-14.00 to 27.10]	8	0.735	0.483	0.02
	BF	91.81 (298.59)	105.57 [-157.82 to 341.44]	7	0.870	0.413	0.03
	VL	33.79 (46.90)	15.63 [-2.26 to 69.84]	8	2.161	0.063	0.2
	GM	-16.30 (175.81)	58.60 [-1541.46 to 118.84]	8	-0.278	0.788	0.01
AL Flat F	TA	19.12 (62.02)	20.67 [-28.55 to 66.80]	8	-0.288	0.925	0.03
	Sol	64.06 (233.04)	77.68 [-115.07 to 243.20]	8	0.825	0.433	0.03
	MG	1.11 (20.91)	6.97 [-14.96 to 17.18]	8	0.159	0.877	0.01
	BF	51.92 (118.07)	41.74 [-46.79 to 150.62]	7	1.244	0.254	0.07
	VL	-3.68 (35.99)	12.00 [-31.34 to 23.98]	8	-0.307	0.767	0.01
	GM	-0.59 (51.23)	17.08 [-39.97 to 38.79]	8	-0.035	0.973	0.01
AL Dec F	TA	21.02 (58.09)	19.36 [-23.63 to 65.67]	8	-0.181	0.086	0.05
	Sol	72.03 (255.94)	85.31 [-124.71 to 268.76]	8	0.844	0.423	0.03
	MG	3.18 (21.65)	7.22 [-19.82 to 13.46]	8	0.440	0.671	0.02
	BF	67.94 (132.23)	46.75 [-42.61 to 178.48]	7	1.453	0.189	0.09
	VL	0.19 (36.46)	12.15 [-27.83 to 28.22]	8	0.016	0.988	0.01
	GM	-27.82 (181.34)	60.45 [-167.21 to 111.57]	8	-0.46	0.658	0.02

DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace, F – Forced Pace, TA- tibialis anterior, Sol- soleus, MG – medial gastrocnemius, BF – biceps femoris, VL – vastus lateralis, GM – gluteus medius

Table H.8 Paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for inter-side comparison of normalised pre-loading activation amplitude difference scores (%) for medial gastrocnemius.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95% CI]	df	t	p	1-β
DL Flat SS	2.67 (24.15)	8.05 [-15.89.23 to 21.23]	8	0.332	0.749	0.01
DL Decline SS	-0.73 (20.03)	6.68 [-16.13 to 14.66]	8	-0.110	0.915	0.01
AL Flat SS	-3.65 (27.99)	9.33 [-25.16 to 17.87]	8	-0.391	0.706	0.01
AL Decline SS	1.29 (24.24)	7.41 [-15.80 to 18.39]	8	0.174	0.866	0.01
DL Flat FP	4.30 (17.19)	5.73 [-8.91 to 17.51]	8	0.75	0.475	0.03
DL Decline FP	1.95 (20.27)	6.76 [-13.64 to 17.53]	8	0.288	0.781	0.01
AL Flat FP	1.06 (17.55)	5.85 [-12.43 to 14.55]	8	0.181	0.861	0.01
AL Decline FP	1.84 (21.97)	7.32 [-15.04 to 18.73]	8	0.252	0.808	0.01

DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace, F – Forced Pace

Appendix I - Statistical results for 3-way ANOVA performed on kinetic derived variables in chapter 3

3-Way ANOVA - $F_{(1,8)} = 0.002$, $p = 0.965$, $(1-\beta = 0.05)$

2-Way Interactions for Pace x Task - $F_{(1, 8)} = 33.11$, $p < 0.01$*, $(1-\beta = 1.0)$

Main effect for Surface - $F_{(1, 8)} = 0.651$, $p = 0.443$, $(1-\beta = 0.11)$

1-way ANOVA for Pace
DL Dec - $F_{(1, 8)} = 11.279$, $p < 0.01$*, $(1-\beta = 0.836)$
AL on Dec - $F_{(1, 8)} = 42.926$, $p < 0.01$*, $(1-\beta = 1.0)$; AL on Flat - $F_{(1, 8)} = 66.052$, $p < 0.01$* $(1-\beta = 1.0)$

1-Way ANOVA for Task
Flat S - $F_{(1, 8)} = 36.626$, $p < 0.01$*, $(1-\beta = 1.0)$; Dec S - $F_{(1, 8)} = 26.399$, $p < 0.01$*, $(1-\beta = 1.0)$

* – $p < 0.05$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-Leg hopping, Flat – Flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced

Figure I.1 Three-way ANOVA on factors for hopping pace, task and surface for hopping frequency.

3-Way ANOVA - $F_{(1,6)} = 1.598$, $p = 0.253$, $(1-\beta = 0.188)$

2-Way Interactions for Task x Surface - $F_{(1, 6)} = 6.0311$, $p < 0.49$, $(1-\beta = 0.539)$

Main effect for Pace - $F_{(1, 6)} = 19.624$, $p < 0.004$*, $(1-\beta = 0.955)$

Post hoc analysis (see table I.1)

1-way ANOVA for Task
Flat S - $F_{(1, 8)} = 25.594$, $p = 0.001$*$(1-\beta = 0.992)$; Flat F - $F_{(1, 6)} = 202.199$, $p < 0.01$*, $(1-\beta = 1.0)$
Dec S - $F_{(1, 6)} = 23.313$, $p < 0.01$*, $(1-\beta = 0.978)$; Dec at F - $F_{(1, 8)} = 33.839$, $p < 0.01$*, $(1-\beta = 1.0)$

1-Way ANOVA for Surface
DL S - $F_{(1, 6)} = 7.111$, $p = 0.03$*, $(1-\beta = 0.648)$

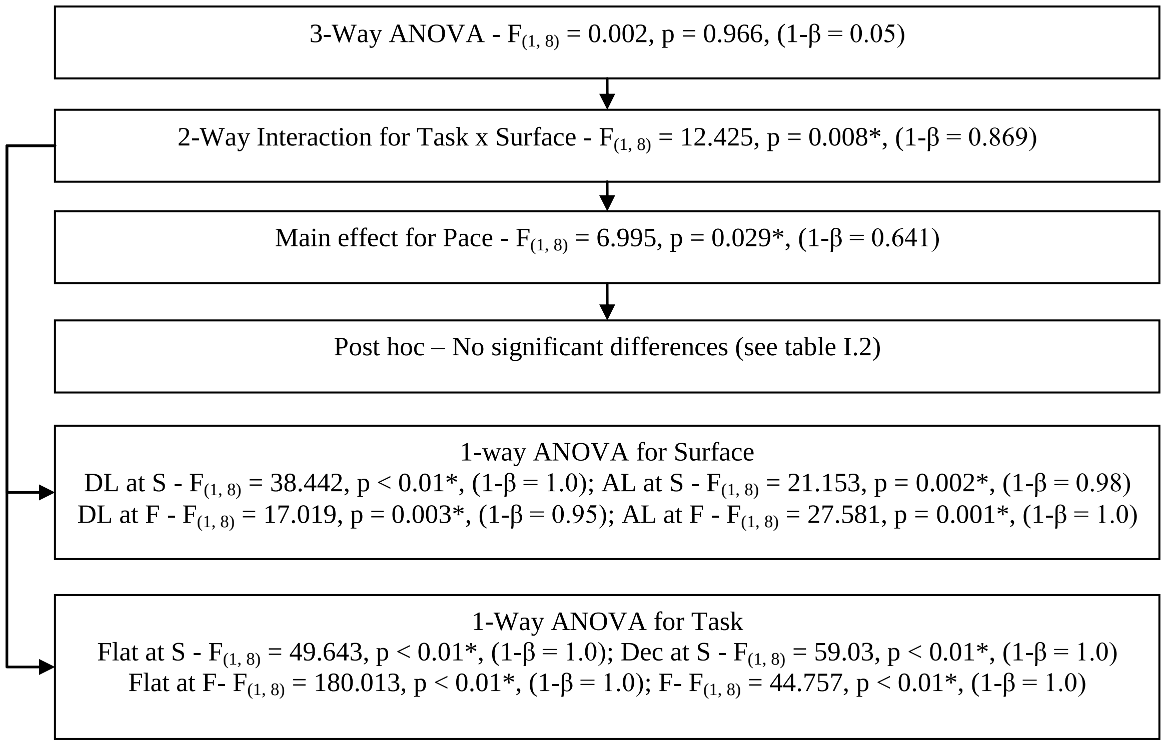
* – $p < 0.05$, † - $p < 0.0125$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL – Alternate-Leg hopping, Flat – Flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure I.2 Three-way ANOVA on factors for hopping pace, task and surface for maximum height in flight phase.

Table I.1 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β) for comparison of vertical height displacement in flight phase (m) between self-selected and forced paces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL Flat	-0.03 (0.03)	0.01 [-0.05 to <0.01]	8	-2.568	0.033	0.47
DL Dec	-0.04 (0.03)	0.01 [-0.06 to -0.01]	8	-3.381	0.010*	0.76
AL Flat	-0.03 (0.02)	0.01 [-0.04 to -0.01]	8	-4.448	0.002*	0.86
AL Dec	-0.02 (0.01)	0.01 [-0.04 to -0.01]	8	-4.138	0.003*	0.99

* - $p < 0.0125$, DL – Double-Leg, AL- Alternate-Leg, Dec – Decline



* – $p < 0.05$, (1- β) – Observed Power, DL - Double-Leg hopping, AL –Alternate-Leg hopping, Flat – Flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure I.3 Three-way ANOVA on factors for hopping pace, task and surface for peak vertical ground reaction force.

Table I.2 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of normalised peak vertical ground reaction force ($N.kg^{-1}$) between self-selected and forced hopping paces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL Flat	-3.06 (3.38)	1.13 [-5.66 to -0.47]	8	-2.719	0.026	0.38
DL Dec	-2.53 (4.40)	1.47 [-5.92 to 0.85]	8	-1.727	0.122	0.14
AL Flat	-1.25 (2.46)	0.82 [-3.14 to 0.64]	8	-1.527	0.165	0.11
AL Dec	-0.64 (2.62)	0.87 [-2.66 to 1.37]	8	-0.733	0.464	0.03

DL- Double-leg, AL – Alternate-leg, Dec – Decline

3-Way ANOVA - $F_{(1, 8)} = 1.686$, $p = 0.23$, ($1-\beta = 0.209$)

↓

No 2-Way Interactions
Pace x Task - $F_{(1, 8)} = 1.649$, $p = 0.235$, ($1-\beta = 0.205$)
Pace x Surface – $F_{(1, 8)} = 0.418$, $p = 0.536$, ($1-\beta = 0.088$)
Task x Surface - $F_{(1, 8)} = 4.26$, $p = 0.073$, ($1-\beta = 0.443$)

↓

Main effects (see table I.3)
Pace - $F_{(1, 8)} = 17.808$, $p = 0.003$*, ($1-\beta = 0.96$)
Task - $F_{(1, 8)} = 20.903$, $p = 0.002$*, ($1-\beta = 0.98$)
Surface - $F_{(1, 8)} = 34.031$, $p < 0.01$*, ($1-\beta = 1.0$)

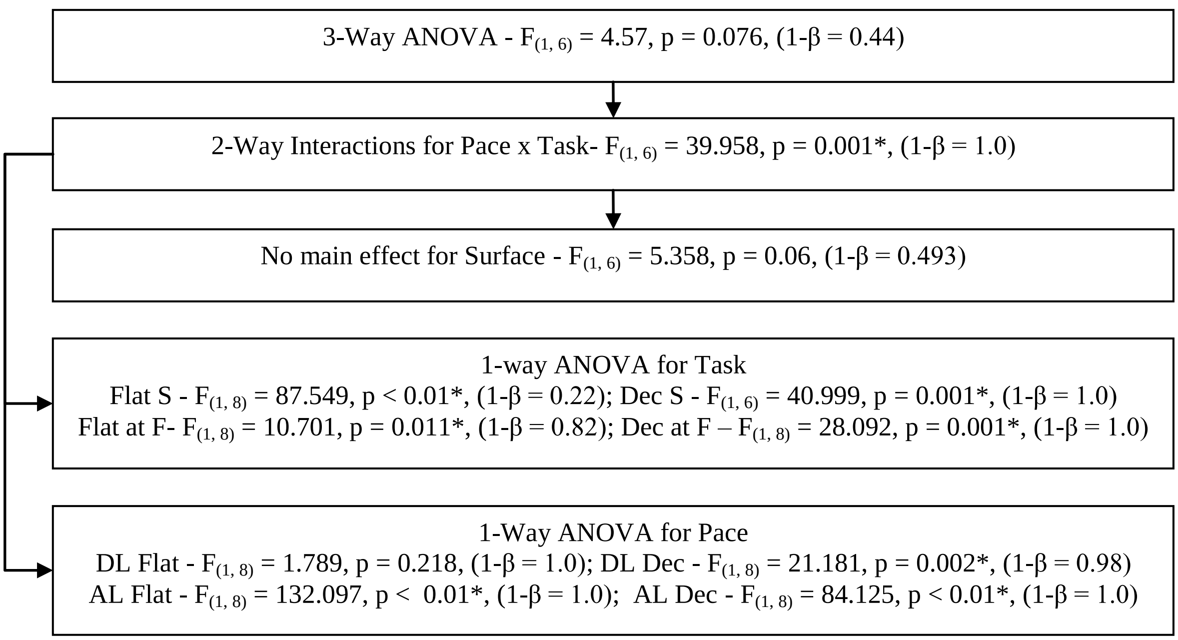
* – $p < 0.05$, (1- β) – Observed Power, DL - Double-Leg hopping, AL –Alternate-Leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure I.4 Three-way ANOVA on factors for hopping pace, task and surface for impulse.

Table I.3 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of impulse (Ns) between paces (self-selected and forced), tasks (double-leg and alternate-leg) and surfaces (flat and decline).

	Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
Pace	DL Flat	-25.97 (126.15)	42.05 [-122.94 to 70.99]	8	-0.618	0.554	0.02
	DL Dec	-86.91 (107.03)	35.68 [-169.18 to -4.64]	8	-2.436	0.041	0.30
	AL Flat	-87.70 (57.73)	19.24 [-132.08 to -43.33]	8	-4.558	0.002*	0.87
	AL Dec	-73.95 (38.69)	12.90 [-103.68 to -44.21]	8	-5.734	≤0.001*	0.98
Task	Flat S	149.76 (88.14)	29.38 [82.01 to 217.51]	8	-2.722	0.001*	0.94
	Dec S	61.84 (125.39)	41.80 [-34.55 to 158.22]	8	-1.479	0.177	0.10
	Flat F	88.03 (68.51)	22.84 [35.37 to 140.69]	8	-1.506	0.005*	0.72
	Dec F	74.80 (54.77)	18.26 [32.70 to 116.9]	8	-0.711	0.003*	0.78
Surface	DL S	167.56 (174.81)	58.27 [33.19 to 301.93]	8	2.876	0.021	0.43
	AL S	79.64 (57.68)	19.23 [35.30 to 123.98]	8	4.142	0.003*	0.79
	DL F	106.63 (50.29)	16.76 [67.97 to 145.29]	8	6.360	≤0.001*	0.99
	AL F	93.40 (40.51)	13.50 [62.26 to 124.53]	8	6.917	≤0.001*	1.0

* - $p < 0.0125$, S – Self-Selected Pace, F – Forced Pace, Dec – Decline, AL – Alternate-Leg, DL – Double-Leg



* – $p < 0.05$, ($1-\beta$) – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure I.5 Three-way ANOVA on factors for hopping pace, task and surface for vertical height displacement during braking phase.

3-Way ANOVA - $F_{(1, 6)} = 0.104$, $p = 0.758$, $(1-\beta = 0.06)$

2-Way Interaction for Pace x Task- $F_{(1, 6)} = 30.549$, $p = 0.002$*, $(1-\beta = 1.0)$

No main effect for Surface - $F_{(1, 6)} = 1.152$, $p = 0.324$, $(1-\beta = 0.15)$

1-way ANOVA for Task
Flat S - $F_{(1, 8)} = 9.134$, $p = 0.017$*, $(1-\beta = 0.75)$
Dec S – $F_{(1, 6)} = 9.804$, $p = 0.02$*, $(1-\beta = 0.74)$

1-Way ANOVA for Pace
DL Dec - $F_{(1, 8)} = 17.863$, $p = 0.003$*, $(1-\beta = 0.96)$
AL Flat - $F_{(1, 8)} = 15.999$, $p = 0.004$*, $(1-\beta = 0.94)$
AL Dec - $F_{(1, 6)} = 20.175$, $p = 0.004$*, $(1-\beta = 0.96)$

* – $p < 0.05$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure I.6 Three-way ANOVA on factors for hopping pace, task and surface for vertical leg stiffness during braking phase.

Appendix J - Statistical results for joint variables in chapter 3

3-Way ANOVA - $F_{(1,8)} = 5.201$, $p = 0.052$, $(1-\beta = 0.52)$

2-Way Interaction

Pace x Task- $F_{(1,8)} = 19.731$, $p = 0.002$*, $(1-\beta = 0.97)$

Pace x Surface – $F_{(1,8)} = 1.151$, $p = 315$, $(1-\beta = 0.16)$

Task x Surface - $F_{(1,8)} = 5.848$, $p = 0.042$*, $(1-\beta = 0.57)$

1-Way ANOVA for Pace

Significant hopping conditions

DL Dec - $F_{(1,8)} = 8.06$, $p =0.022$*, $(1-\beta = 0.70)$

AL Flat - $F_{(1,8)} = 34.593$, $p < 0.001$*, $(1-\beta = 1.0)$

AL Dec- $F_{(1,8)} = 43.456$, $p < 0.001$*, $(1-\beta = 1.0)$

1-Way ANOVA for Task

Significant hopping conditions

Flat S - $F_{(1,8)} = 12.311$, $p = 0.008$*, $(1-\beta = 0.87)$

Dec S - $F_{(1,8)} = 6.656$, $p = 0.033$*, $(1-\beta = 0.62)$

1-Way ANOVA for Surface

Significant hopping conditions

DL S - $F_{(1,8)} = 11.255$, $p = 0.01$*, $(1-\beta = 0.84)$

AL S - $F_{(1,8)} = 31.945$, $p < 0.001$*, $(1-\beta = 1.0)$

AL F - $F_{(1,8)} = 46.552$, $p < 0.001$*, $(1-\beta = 1.0)$

* – $p < 0.05$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.1 Three-way ANOVA on factors for hopping pace, task and surface for ankle joint excursion during braking phase.

3-Way ANOVA - $F_{(1,8)} = 0.022$, $p = 0.886$, ($1-\beta = 0.052$)

2-Way Interaction
Pace x Task- $F_{(1,8)} = 11.708$, $p = 0.009$*, ($1-\beta = 0.85$)
Pace x Surface – $F_{(1,8)} = 0.041$, $p = 0.844$, ($1-\beta = 0.054$)
Task x Surface - $F_{(1,8)} = 3.583$, $p = 0.095$, ($1-\beta = 0.385$)

1-Way ANOVA for Pace

Hopping conditions
DL Flat - $F_{(1,8)} = 1.119$, $p = 0.305$, ($1-\beta = 0.36$)
DL Dec - $F_{(1,8)} = 1.942$, $p = 0.201$, ($1-\beta = 0.233$)
AL Flat - $F_{(1,8)} = 21.905$, $p < 0.002$*, ($1-\beta = 0.98$)
AL Dec- $F_{(1,8)} = 12.171$, $p < 0.008$*, ($1-\beta = 0.86$)

1-Way ANOVA for Task

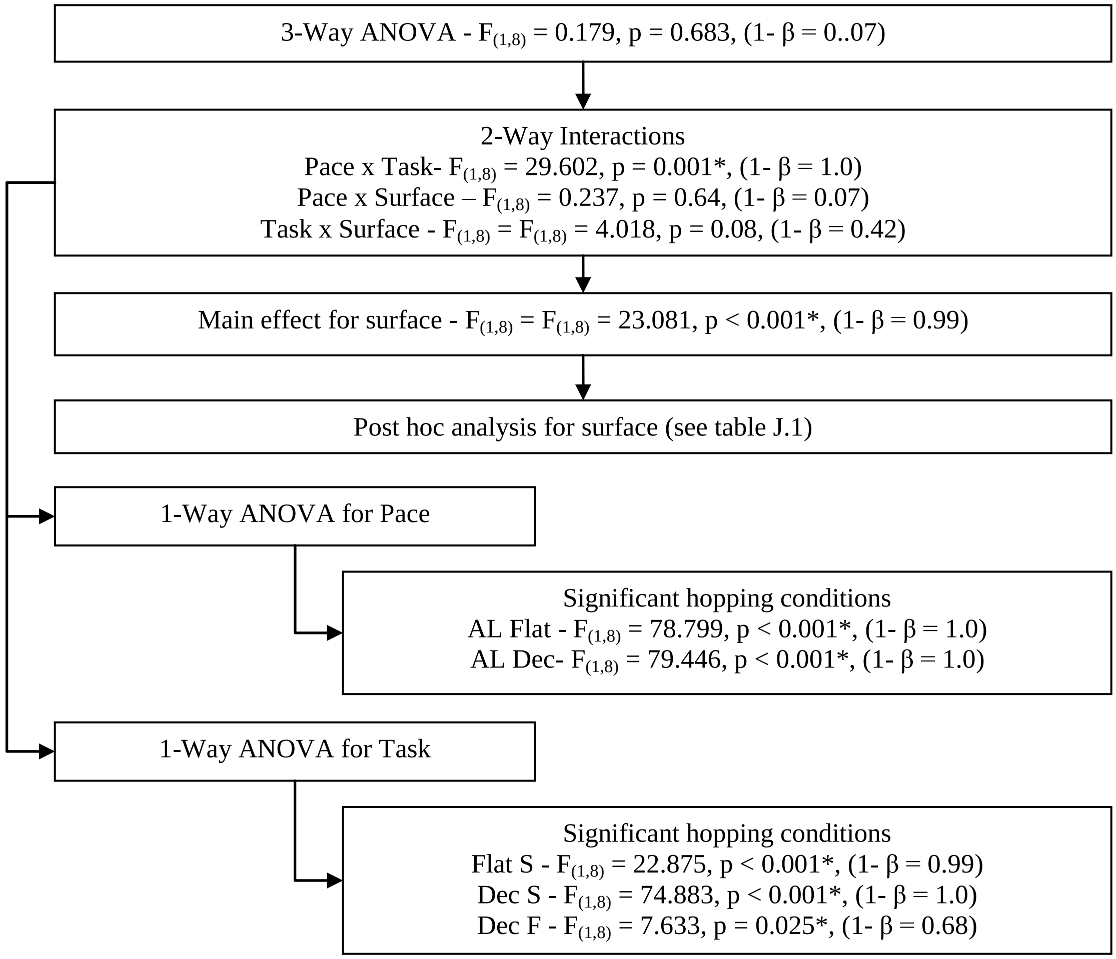
Hopping conditions
Flat S - $F_{(1,8)} = 16.549$, $p = 0.004$*, ($1-\beta = 0.94$)
Flat F - $F_{(1,8)} = 6.048$, $p = 0.035$*, ($1-\beta = 0.61$)
Dec S - $F_{(1,8)} = 27.102$, $p < 0.001$*, ($1-\beta = 1.0$)
Dec F - $F_{(1,8)} = 21.348$, $p < 0.002$*, ($1-\beta = 0.98$)

1-Way ANOVA for Surface

Hopping conditions
DL S - $F_{(1,8)} = 37.236$, $p < 0.001$*, ($1-\beta = 1.0$)
DL F - $F_{(1,8)} = 19.305$, $p < 0.002$*, ($1-\beta = 0.97$)
AL S - $F_{(1,8)} = 4.998$, $p = 0.056$, ($1-\beta = 0.502$)
AL F - $F_{(1,8)} = 9.595$, $p = 0.015$*, ($1-\beta = 0.774$)

* – $p < 0.05$, ($1-\beta$) – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – Flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.2 Three-way ANOVA on factors for hopping pace, task and surface for knee joint excursion during braking phase.



* – $p < 0.05$, ($1- \beta$) – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.3 Three-way ANOVA on factors for hopping pace, task and surface for hip joint excursion during braking phase.

Table J.1 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of hip joint excursion (degrees) between flat and decline surfaces for hopping.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL S	-5.78 (4.71)	1.57 [-9.40 to -2.16]	8	-3.682	0.006*	0.68
AL S	-3.08 (4.19)	1.40 [-6.30 to 0.14]	8	-2.209	0.058	0.24
DL F	-5.89 (4.61)	1.54 [-9.44 to -2.35]	8	-3.831	0.005*	0.72
AL F	-4.11 (2.79)	0.93 [-6.26 to -1.96]	8	-4.416	0.002*	0.85

* - $p < 0.0125$, DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace; F – Forced Pace

3-Way ANOVA
$F_{(1, 8)} = 0.931$, $p = 0.363$, $(1-\beta = 0.14)$

↓

2-Way Interactions
Pace x Surface - $F_{(1, 8)} = 0.469$, $p = 0.513$, $(1-\beta = 0.09)$
Task x Surface - $F_{(1, 8)} = 0.701$, $p = 0.427$, $(1-\beta = 0.12)$
Pace x Task - $F_{(1, 8)} = 0.854$, $p = 0.382$, $(1-\beta = 0.14)$

↓

Main Effects
Task - $F_{(1, 8)} = 20.222$, $p = 0.002$*, $(1-\beta = 0.98)$
Surface - $F_{(1, 8)} = 63.26$, $p < 0.001$*, $(1-\beta = 1.0)$

↓

Post hoc analysis
(see table J.2 for task and J.3 for surface)

* – $p < 0.05$, (1- β) – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – Flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.4 Three-way ANOVA on factors for hopping pace, task and surface ankle joint moment during braking phase.

Table J.2 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of ankle joint moment ($Nm.kg^{-1}$) between double-leg and alternate-leg hopping tasks.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
Flat S	-0.74 (0.79)	0.26 [-1.35 to -0.13]	8	-2.800	0.023	0.41
Dec S	-0.44 (0.33)	0.11 [-0.69 to -0.19]	8	-4.039	*0.004	0.76
Flat F	-0.47 (0.53)	0.18 [-0.88 to -0.06]	8	-2.623	0.031	0.37
Dec F	-0.43 (0.46)	0.15 [-0.78 to -0.08]	8	-2.842	0.022	0.41

* - $p < 0.0125$, S- Self-Selected Pace, F – Forced Pace, Dec - Decline

Table J.3 Post hoc paired t-test results mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of ankle joint moment ($Nm.kg^{-1}$) between flat and decline hopping surfaces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL S	0.87 (0.44)	0.15 [0.54 to 1.21]	8	5.994	*≤0.001	0.99
AL S	1.18 (0.43)	0.14 [0.85 to 1.50]	8	8.266	*≤0.001	1.0
DL F	1.13 (0.81)	0.27 [0.50 to 1.75]	8	4.166	*0.003	0.80
AL F	1.16 (0.68)	0.23 [0.64 to 1.69]	8	5.138	*0.001	0.94

* - $p < 0.0125$, 1-β – Observed Power, DL – Double-Leg, AL – Alternate-leg, S- Self-Selected Pace, F – Forced Pace

3-Way ANOVA
$F_{(1, 8)} = 1.114$, $p = 0.322$, ($1-\beta = 0.15$)

↓

2-Way Interactions
Pace x Surface - $F_{(1, 8)} = 3.736$, $p = 0.089$, ($1-\beta = 0.398$)
Task x Surface - $F_{(1, 8)} = 0.375$, $p = 0.557$, ($1-\beta = 0.084$)
Pace x Task - $F_{(1, 8)} = 4.071$, $p = 0.078$, ($1-\beta = 0.427$)

↓

Main Effects
Task - $F_{(1, 8)} = 10.512$, $p = 0.012$*, ($1-\beta = 0.81$)
Pace - $F_{(1, 8)} = 17.188$, $p = 0.003$*, ($1-\beta = 0.95$)

↓

Post hoc analysis
(see table J.4 for task and J.5 for surface)

* – $p < 0.05$, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.5 Three-way ANOVA on factors for hopping pace, task and surface for the change in knee joint moment during braking phase.

Table J.4 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of knee joint moment ($Nm.kg^{-1}$) between double-leg and alternate-leg hopping tasks.

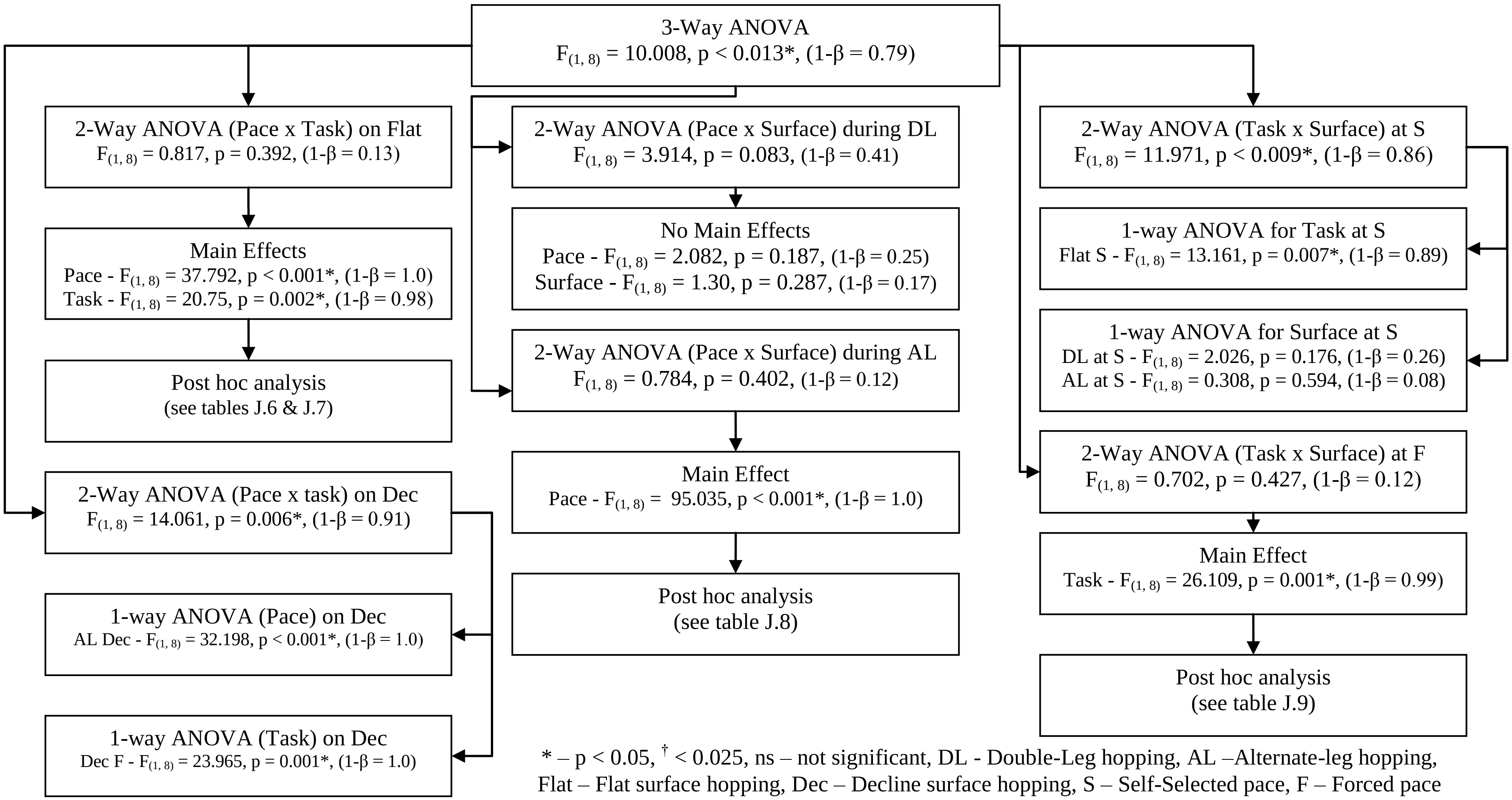
Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
Flat S	0.69 (0.54)	0.18 [0.27 to 1.1]	8	3.833	0.005*	0.71
Decline S	0.42 (0.39)	0.13 [0.12 to 0.72]	8	3.263	0.011*	0.54
Flat F	0.19 (0.80)	0.26 [-0.42 to 0.80]	8	0.728	0.487	0.03
Decline F	0.23 (0.43)	0.14 [-0.10 to 0.57]	8	1.613	0.145	0.12

* - p < 0.0125, DL – Double-Leg, AL – Alternate-leg, S- Self-Selected Pace, F – Forced Pace

Table J.5 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of knee joint moment ($Nm.kg^{-1}$) between self-selected and forced hopping paces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL flat	-0.32 (0.71)	0.24 [-0.87 to 0.22]	8	-1.359	0.211	0.08
DL Dec	-0.28 (0.41)	0.14 [-0.60 to 0.04]	8	-2.000	0.081	0.20
AL Flat	-0.82 (0.42)	0.14 [-1.14 to -0.50]	8	-5.824	≤0.001*	0.98
AL Dec	-0.46 (0.39)	0.13 [-0.76 to -0.161]	8	-3.548	0.008*	0.64

* - p < 0.0125, DL – Double-Leg, AL – Alternate-leg, S- Self-Selected Pace, F – Forced Pace



* – $p < 0.05$, † < 0.025, ns – not significant, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – Flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.6 Three-way ANOVA on factors for hopping pace, task and surface for the change in hip joint moment during braking phase.

Table J.6 Post hoc paired t-test results mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of hip joint moment ($Nm.kg^{-1}$) between self-selected and forced hopping paces on a flat surface.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL Flat	-0.37 (0.45)	0.15 [-0.71 to -0.02]	8	-2.462	0.039	0.44
AL Flat	-0.53 (0.18)	0.06 [-0.67 to -0.39]	8	-8.660	*≤0.001	1.0

* - p < 0.025, DL – Double-Leg, AL – Alternate-Leg

Table J.7 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of hip joint moment ($Nm.kg^{-1}$) between double-leg and alternate-leg hopping tasks on a flat surface.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
Flat S	-0.75 (0.62)	0.21 [-1.22 to -0.27]	8	-3.628	*0.007	0.79
Flat F	-0.90 (0.59)	0.20 [-1.36 to -0.45]	8	-4.596	*0.002	0.94

* - p < 0.025, S- Self-Selected Pace, F – Forced Pace

Table J.8 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of hip joint moment ($Nm.kg^{-1}$) between self-selected and forced hopping paces during AL hopping.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
AL Flat	-0.53 (0.18)	0.060 [-0.66 to -0.39]	8	-8.660	*≤0.001	1.0
AL Dec	-0.65 (0.34)	0.11 [-0.91 to -0.38]	8	-5.674	*≤0.001	1.0

* - p < 0.025, AL – Alternate-Leg, Dec - Decline

Table J.9 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of hip joint moment ($Nm.kg^{-1}$) between double-leg and alternate-leg hopping tasks at a forced pace.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
Flat F	-0.90 (0.59)	-1.36 [-0.67 to -0.45]	8	-4.596	*0.002	0.94
Dec F	-0.79 (0.48)	0.16 [-1.16 to -0.42]	8	-4.895	*0.001	0.97

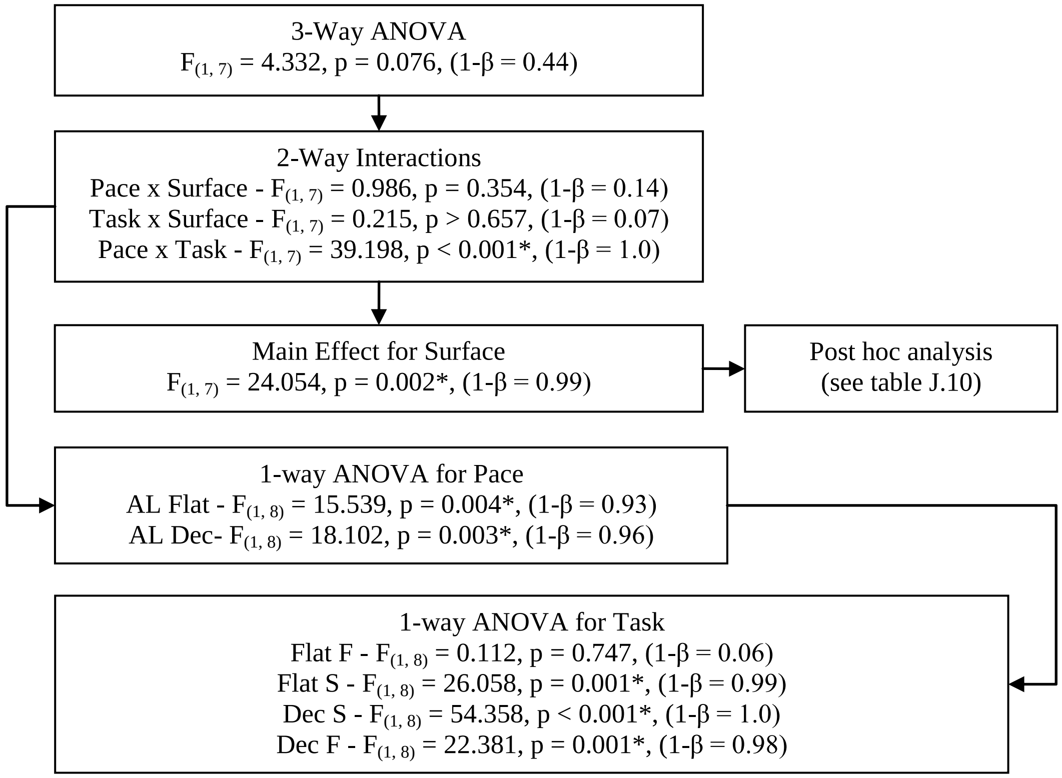
* - p < 0.025, F – Forced Pace, Dec - Decline

Appendix J.7.Supplement - Analysis of Ankle Joint Stiffness

Ankle joint stiffness for all nine participants during DL hopping on a decline surface at a self-selected pace was mean 4.94Nm.kg^{-1} (SD) (4.77) with a CV of 96.5%. A single participant had a value of ankle joint stiffness (17.56 Nm.kg^{-1}.rad^{-1}) that was greater than two standard deviations from the mean for the group. Removing the single sample to reduce the sample to eight reduced the CV to 19.1%, similar to the range of CV for ankle joint stiffness across the other hopping conditions (13% to 37%). Hence, prior to data analysis of ankle joint stiffness the single outlier was removed to reduce the risk of a type II error.

On analysis of this single participant, right ankle excursion was found to be low (5.2º) and less than two standard deviations from the mean for pooled data of all nine participants. Joint moments on both sides for this participant were within range (1.55 to 2.81Nm.kg^{-1}) of all samples from both the left and right sides recorded during this hopping condition. Thus, the calculated ankle joint stiffness was greater than two standard deviations from the mean of pooled data for all nine participants. The sample was removed to reduce the risk of a type II error.

Although the ankle excursion was small it was not removed from analysis of ankle excursion (Figure J.1) as the CV for pooled data of all 9 samples (20.3%) approximated the CV across all other hopping conditions (9 to 20.4%). Thus the ankle excursion was considered representative of the sample population.



* – $p < 0.05$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.7 Three-way ANOVA on factors for hopping pace, task and surface for ankle joint stiffness during braking phase.

Table J.10 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of ankle joint stiffness ($Nm.kg^{-1}.rad^{-1}$) between flat and decline surfaces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL S	0.75 (0.51)	0.18 [-0.32 to 1.18]	7	4.151	*0.004	0.76
AL S	1.18 (0.89)	0.30 [0.50 to 1.86]	8	3.993	*0.004	0.75
DL F	1.74 (1.77)	0.59 [0.38 to 3.10]	8	2.944	0.019	0.45
AL F	0.95 (0.97)	0.32 [0.21 to 1.70]	8	2.945	0.019	0.45

* - $p < 0.0125$, DL – Double-Leg, AL – Alternate-leg, S- Self-Selected Pace, F – Forced Pace

3-Way ANOVA $F_{(1, 8)} = 1.002$, $p = 0.346$, $(1-\beta = 0.14)$

↓

No 2-Way Interactions
Pace x Surface - $F_{(1, 8)} = 0.43$, $p = 0.53$, $(1-\beta = 0.09)$
Task x Surface - $F_{(1, 8)} = 0.325$, $p = 0.564$, $(1-\beta = 0.08)$
Pace x Task - $F_{(1, 8)} = 3.363$, $p = 0.104$, $(1-\beta = 0.37)$

↓

No Main Effects
Pace- $F_{(1, 8)} = 1.925$, $p = 0.203$, $(1-\beta = 0.23)$
Task – $F_{(1, 8)} = 2.292$, $p = 0.168$, $(1-\beta = 0.27)$
Surface - $F_{(1, 8)} = 2.274$, $p = 0.136$, $(1-\beta = 0.31)$

* – $p < 0.05$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.8 Three-way ANOVA on factors for hopping pace, task and surface knee joint stiffness during braking phase.

Appendix J.8.Supplement - Analysis of knee joint stiffness.

Joint stiffness was derived by dividing the joint moment (Nm) by angular displacement (radians) during braking phase. In braking phase it was assumed that the knee would only flex. Knee joint moment and knee joint excursion were expressed as positive values during braking phase as the knee joint flexes to absorb kinetic energy. Figure 2.4.2.2.8 illustrates the standard deviation to include values of negative knee joint stiffness. A negative knee joint stiffness may be resultant due to a number of factors where it is unexpected. The error in knee moment may be due to random error associated with the sampling of motion and/or following the filtering procedure.

Either the numerator (joint moment) or denominator (angular displacement) may be negative. During AL hopping on a flat surface at a self-selected pace, the finding was that 15 hops on each side, recorded from a total of 90 hops on each side for all participants, had negative knee moments. All 10 consecutive hops for one participant on the left side were negative. Thus, 16.7% of the total hops recorded had negative knee joint moments during AL hopping on a flat surface at a flat self-selected pace. To determine whether negative knee joint moments influenced the finding of possible negative knee joint stiffness all negative hops were removed from analysis. Knee joint moment and knee joint excursion were calculated on each side for hops with a positive knee joint moments only. Data were compared across the two procedures with the first procedure including all hops and the second procedure including hops with only a positive knee joint moment. Multiple paired t-tests were performed with significance accepted at $p < 0.05$ (PASW Statistics, version 18 and G*Power 3.1.2, http://www.psycho.uni-duesseldorf.de). The results are presented in tables J.8.S.1 and J.8.S.2.

Table J.8.S.1 Knee joint moment (kNm.kg^{-1}), joint excursions (radians) and joint stiffness (kNm.kg^{-}1.rad^{-1}) (mean (SD) [CV]) during braking phase calculated using the two procedures.

Derived variable	All Hops (180)		Hops with positive knee moments only (150)	
	Left	Right	Left	Right
No. of hops	90	90	75	75
Moment	0.61	0.71	0.80	0.79
(Nm.kg^{-1})	(0.74)	(0.75)	(0.65)	(0.72)
	[0.12]	[0.11]	[0.08]	[0.09]
Excursion	0.18	0.12	0.19	0.12
(radians)	(0.08)	(0.08)	(0.09)	(0.09)
	[45.7]	[67.0]	[45.8]	[73.8]
Joint stiffness	3.46	14.15	4.49	16.29
(kNm.kg^{-1}.rad^{-1})	(3.55)	(20.84)	(2.84)	(20.98)
	[0.10]	[0.15]	[0.06]	[0.13]

Table J.8.S.2 Results of multiple paired t-tests (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) comparing derived variables of knee moment (M) (kNm.kg^{-1}), knee excursion (E) (radians) and knee joint stiffness (K) (kNm.kg^{-1}.rad^{-1}) on left and right sides across two procedures.

		Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
M	L	-0.09 (0.16)	0.06 [-0.22 to 0.05]	7	-1.516	0.173	0.11
	R	-0.09 (0.12)	0.04 [-0.17 to 0.004]	8	-2.212	0.058	0.25
E	L	-0.002 (0.007)	0.002 [-0.007 to 0.003]	8	-0.964	0.363	0.04
	R	-0.005 (0.006)	0.006 [-0.02 to 0.008]	8	-0.914	0.388	0.32
K	L	-0.53 (0.77)	0.26 [-1.12 to 0.07]	8	-2.052	0.074	0.21
	R	-0.94 (2.27)	0.76 [-2.68 to 0.81]	8	-1.237	0.251	0.07

M – Knee joint Moment; E – Knee joint Excursion, K - Knee joint Stiffness, L – Left, R – Right

The key finding was that there was no significant difference in knee joint moment, excursion or stiffness between the two procedures. Thus, accounting for recorded negative knee moments did not influence knee joint mechanics on the left or right significantly.

The CV for all three derived variables was reduced using hops with positive knee moments only; however on the right side the knee joint stiffness still had a CV of 128.8%. This large variance as compared to the left side may indicate that there may be outliers in the data set of nine participants for the measure of right knee joint stiffness. Two of nine participants have right knee joint stiffness values that are each less than and greater than the 95% CI (1.02 kNm.kg^{-}1.rad^{-1} to 12.72 kNm.kg^{-}1.rad^{-1}) across all hopping conditions. These values are 0.49kNm.kg^{-}1.rad^{-1} and 61.73kNm.kg^{-}1.rad^{-1}. For the lowest value, the measure of knee moment is small (0.11 $kNm.kg^{-1}$) and for the highest value, the measure of knee excursion is small (0.5º or 0.009 radians). Hence, measures of right knee joint stiffness during AL hopping on the flat surface at a self-selected pace has a large variance with outliers both greater and less than two standard deviations from the mean of pooled data for all nine samples.

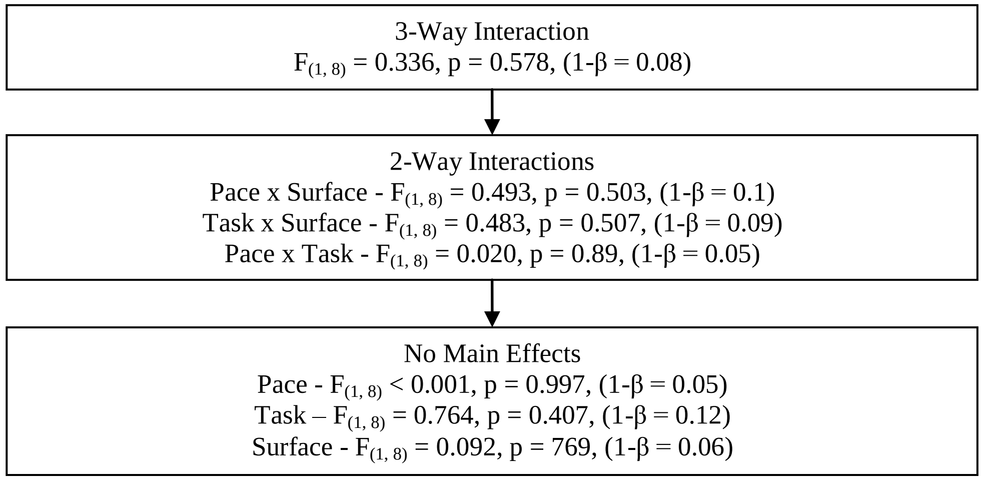
Statistical analysis was performed to compare inter-side difference of knee joint stiffness during AL on a flat surface at a self-selected pace to assess whether data could be pooled across sides and not be influenced by a greater mean and variance on the right than the left. Thus, the three conditions for data inclusion were; using all hops (a), including hops with only positive knee moments (b) or removing the two outliers from the sample of hops with only positive knee moments (c). There was no statistical inter-side difference in knee joint stiffness under a, b or c (Table J.8.S.3). Although removing the two outliers from the data set reduced the CV from 127.5% to 77.0%. With no significant inter-side difference, data were pooled between sides and a 3-way ANOVA performed for comparison of knee joint stiffness across hopping conditions (PASW Statistics, version 18 and G*Power 3.1.2, http://www.psycho.uni-duesseldorf.de). All hops were included in the 3-way ANOVA as there was no difference in derived variables whether all hops were included or only those with a positive knee moment and as there was not a large reduction in the CV on removing two samples (Table J.8.S.3). All nine samples were included to reduce the risk of a type I error.

Table J.8.S.3 Results of multiple paired t-tests (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value and p value) comparing left and right sides for knee joint stiffness ($kNm.kg^{-1}.rad^{-1}$) across all hops (a), hops with a positive knee moment only (b) and hops with a positive knee moment with two outliers removed (c).

	Mean Difference (SD)	Standard error of the mean [95%CI of the difference]	df	t	p	1-β
a	-10.69 (19.73)	6.58 [-25.86 to 4.48]	8	1.625	0.143	0.48
b	-13.23 (21.83)	8.25 [-33.42 to 6.96]	6	-1.604	0.160	0.48
c	-7.24 (12.20)	5.46 [-22.39 to 7.91]	4	-1.326	0.108	0.26

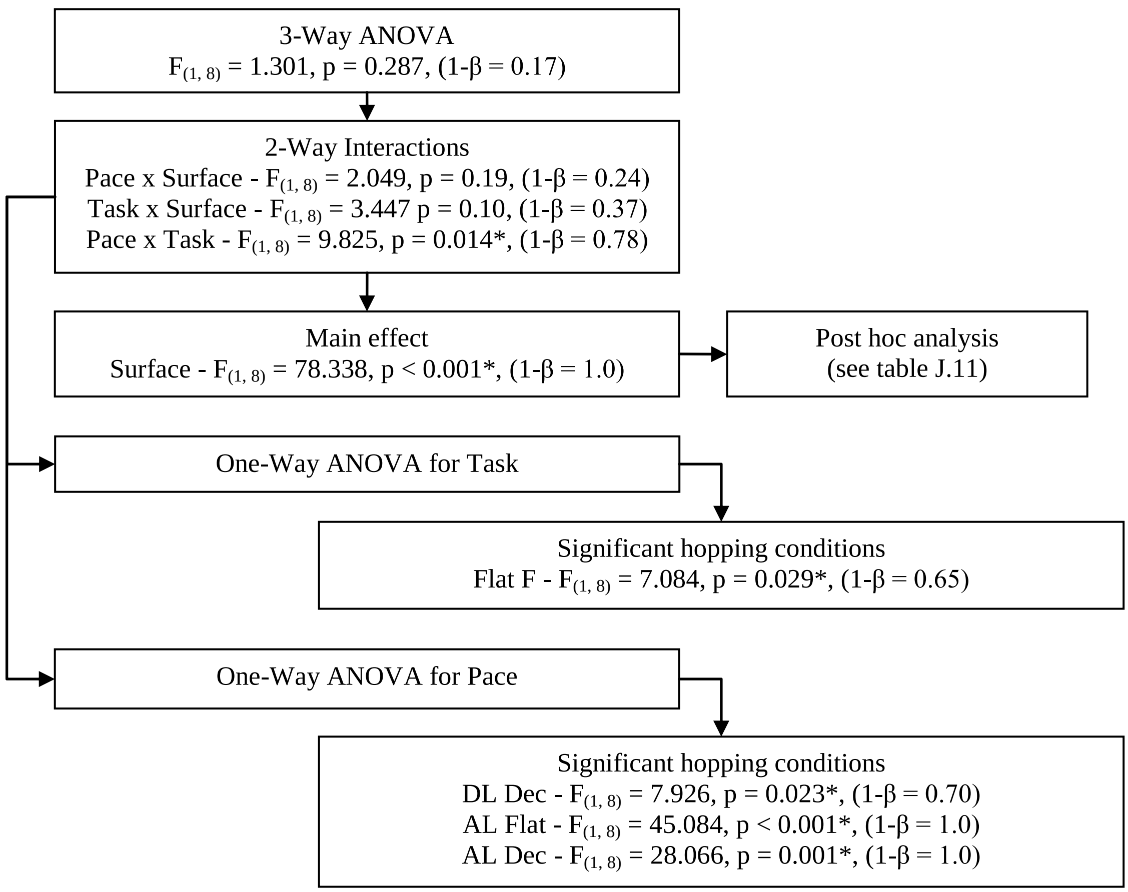
a – all hops, b – hops with positive knee moments, c – hops with positive knee moments and two outliers removed

However, comparing joint moments for pooled data of all 180 hops for this hopping condition to pooled data that included joint moments that were positive only did not significantly change the pooled mean and standard deviation of knee joint moment on either side. There were no recordings of negative angular displacements at the knee on either side. It is suggested that the large variance in joint stiffness is a result of a small sample size.



* – $p < 0.05$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.9 Three-way ANOVA on factors for hopping pace, task and surface for the change in hip joint stiffness during braking phase.



* – $p < 0.05$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.10 Three-way ANOVA on factors for hopping pace, task and surface for ankle joint work during braking phase.

Table J.11 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of mechanical work done (J.kg^{-1}) in the braking phase at the ankle joint between flat and decline hopping surfaces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL S	0.83 (0.48)	0.16 [0.46 to 1.20]	8	5.137	0.001*	0.95
AL S	0.89 (0.31)	0.10 [0.65 to 1.13]	8	8.672	≤0.001*	1.0
DL F	0.78 (0.50)	0.17 [0.39 to 1.16]	8	4.672	0.002*	0.89
AL F	1.21 (0.54)	0.18 [0.79 to 1.63]	8	6.692	≤0.001*	1.0

* - p < 0.0125, DL – Double-Leg, AL – Alternate-leg, S- Self-Selected Pace, F – Forced Pace

3-Way ANOVA
$F_{(1, 8)} = 0.329$, $p = 0.582$, $(1-\beta = 0.08)$

↓

2-Way Interactions
Pace x Surface - $F_{(1, 8)} = 2.651$, $p = 0.142$, $(1-\beta = 0.30)$
Task x Surface - $F_{(1, 8)} = 3.325$ $p = 0.106$, $(1-\beta = 0.36)$
Pace x Task - $F_{(1, 8)} = 3.667$, $p = 0.092$, $(1-\beta = 0.39)$

↓

Main effects
Pace - $F_{(1, 8)} = 23.783$, $p = 0.001$*, $(1-\beta = 0.99)$
Task – $F_{(1, 8)} = 35.534$, $p < 0.001$*, $(1-\beta = 1.0)$
Surface - $F_{(1, 8)} = 17.568$, $p = 0.003$*, $(1-\beta = 0.96)$

↓

Post hoc analysis
(see tables J.12, I.13 and J.14)

* – p < 0.05, (1-β) – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.11 Three-way ANOVA on factors for hopping pace, task and surface for knee joint work during braking phase.

Table J.12 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of mechanical work done ($J.kg^{-1}$) in the braking phase at the knee joint between self-selected and forced hopping paces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL Flat	-0.13 (0.43)	0.16 [-0.05 to 0.24]	8	-0.787	0.454	0.04
DL Dec	-0.26 (0.28)	0.095 [-0.48 to -0.04]	8	-2.708	0.027	0.40
AL Flat	-0.42 (0.20)	0.07 [-0.58 to -0.27]	8	-6.442	≤0.001*	0.99
AL Dec	-0.46 (0.20)	0.07 [-0.61 to -0.31]	8	-7.038	≤0.001*	1.0

* - $p < 0.0125$, DL – Double-Leg, AL – Alternate-leg, Dec - Decline

Table J.13 Post hoc paired t-test results (degrees of freedom, t and p values) and mean difference (S), standard error [95%CI], degrees of freedom (df), t value, p value and observed power (1-β) for comparison of mechanical work done ($J.kg^{-1}$) in the braking phase at the knee joint between double-leg and alternate-leg hopping tasks.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
Flat S	0.40 (0.24)	0.08 [0.21 to 0.59]	8	4.873	0.001*	0.93
Dec S	0.50 (0.23)	0.08 [0.31 to 0.67]	8	6.377	≤0.001*	1.0
Flat F	0.10 (0.44)	0.15 [-0.24 to 0.44]	8	0.677	0.518	0.03
†Flat F	0.23 (0.21)	0.07 [0.06 to 0.40]	7	3.194	0.015	0.47
Dec F	0.29 (0.26)	0.09 [0.09 to 0.49]	8	3.352	0.010*	0.58

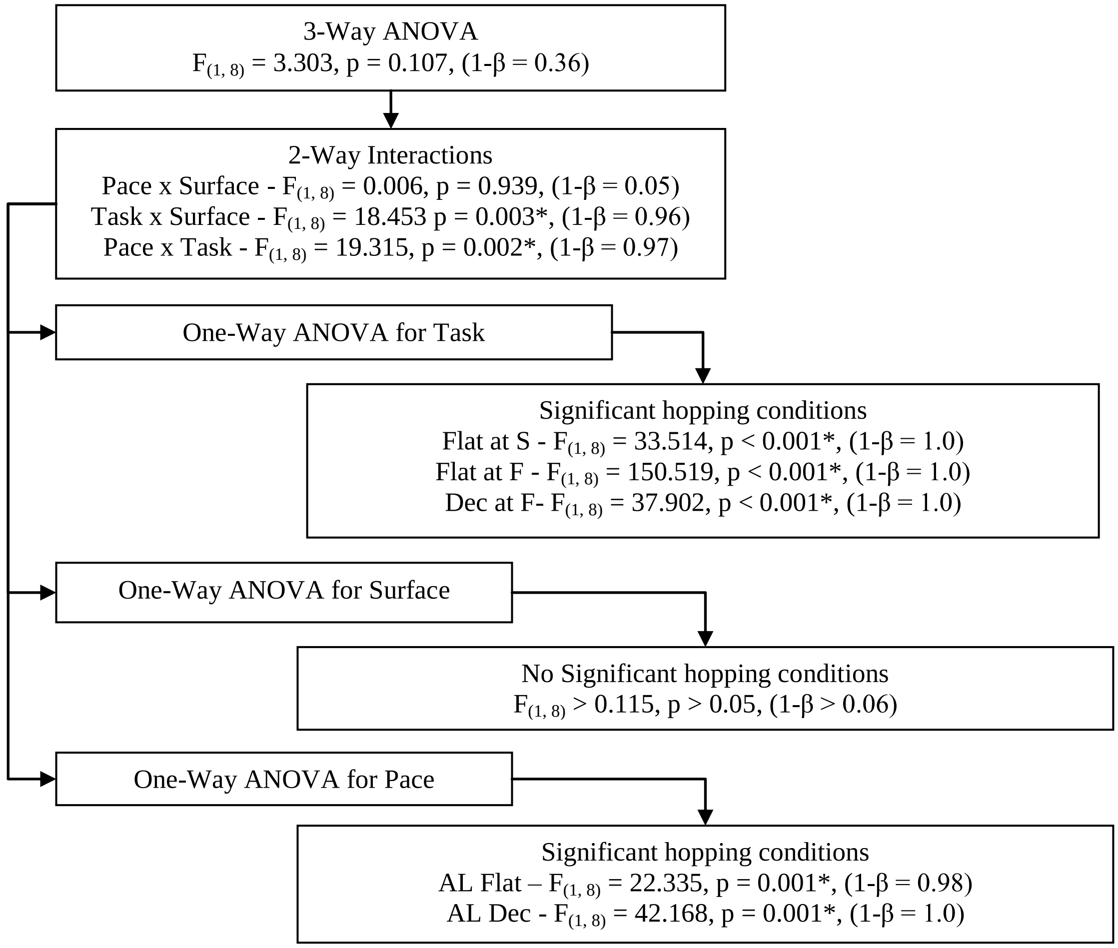
* - $p < 0.0125$, Dec – Decline, S – Self-Selected Pace, F – Forced Pace

† - df reduced by removing one sample from total of nine as the CV was large

Table J.14 Post hoc paired t-test results (degrees of freedom, t and p values) and mean difference (S), standard error [95%CI], degrees of freedom (df), t value, p value and observed power (1-β) for comparison of mechanical work done ($J.kg^{-1}$) in the braking phase at the knee joint between flat and decline hopping surfaces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL S	-0.28 (0.32)	0.11 [-0.52 to -0.04]	8	-2.650	0.029	0.36
AL S	-0.19 (0.20)	0.07 [-0.34 to -0.03]	8	-2.732	0.026	0.42
DL F	-0.41 (0.25)	0.08 [-0.60 to -0.22]	8	-4.861	0.001*	0.92
AL F	-0.22 (0.30)	0.10 [-0.45 to 0.001]	8	-2.215	0.058	0.24

* - $p < 0.0125$, DL – Double-Leg, AL – Alternate-leg, S- Self-Selected Pace, F – Forced Pace



* – $p < 0.05$, $(1-\beta)$ –Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – flat surface hopping, Dec – Decline surface hopping, S – Self-Selected pace, F – Forced pace

Figure J.12 Three-way ANOVA on factors for hopping pace, task and surface for hip joint work during braking phase.

Appendix K - Statistical results for electromyography derived variables in chapter 3

Left TA 3-Way ANOVA - $F_{(1, 8)} = 0.001$, $p = 0.982$, ($1\text{-}\beta = 0.05$)

No 2-Way Interactions

Pace x Task – $F_{(1, 8)} = 0.344$, p 0.574, ($1\text{-}\beta = 0.08$)

Task Surface - $F_{(1, 8)} = 0.041$, $p = 0.844$, ($1\text{-}\beta = 0.26$)

Pace x Surface - $F_{(1, 8)} = 2.264$, $p = 0.171$, ($1\text{-}\beta = 0.05$)

No Main Effects

Pace - $F_{(1, 8)} = 1.057$, $p = 0.334$, ($1\text{-}\beta = 0.15$)

Task - $F_{(1, 8)} = 1.959$, $p = 0.199$, ($1\text{-}\beta = 0.24$)

Surface - $F_{(1, 8)} = 5.294$, $p = 0.05$, ($1\text{-}\beta = 0.53$)

Right TA - 3-Way ANOVA - $F_{(1, 8)} = 0.01$, $p = 0.922$, ($1\text{-}\beta = 0.05$)

2-Way Interaction(s)

Task x Surface - $F_{(1, 8)} = 5.829$, $p = 0.042$*, ($1\text{-}\beta = 0.56$)

No main effect for Pace - $F_{(1, 8)} = 0.815$, $p = 0.393$, ($1\text{-}\beta = 0.13$)

One-way ANOVA for Task - Flat F - $F_{(1, 8)} = 7.993$, $p = 0.022$*, ($1\text{-}\beta = 0.70$)

One– way ANOVA for Surface – DL F - $F_{(1, 8)} = 5.39$, $p = 0.049$*, ($1\text{-}\beta = 0.53$)

TA – tibialis anterior, * – $p < 0.05$, ($1\text{-}\beta$) – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – Flat hopping surface, Dec – Decline hopping surface S – Self-Selected Pace, F – Forced pace

Figure K.1 Three-way ANOVA for left and right tibialis anterior (TA) for the derived variable of peak activation amplitude.

Left Sol 3-Way ANOVA - $F_{(1, 8)} = 0.108$, $p = 0.751$, ($1\text{-}\beta = 0.06$)

No 2-Way Interactions
Pace x Task - $F_{(1, 8)} = 2.121$, $p = 0.183$, ($1\text{-}\beta = 0.25$)
Task x Surface– $F_{(1, 8)} = 0.504$, $p = 0.498$, ($1\text{-}\beta = 0.10$)
Pace x Surface- $F_{(1, 8)} = 3.869$, $p = 0.085$, ($1\text{-}\beta = 0.41$)

Main effect for Surface - $F_{(1, 8)} = 6.038$, $p = 0.039$*, ($1\text{-}\beta = 0.58$)

Post hoc – no significant differences (see table K.1)

Right Sol - 3-Way ANOVA - $F_{(1, 7)} = 2.331$, $p = 0.171$, ($1\text{-}\beta = 0.26$)

No 2-Way Interactions
Pace x Task - $F_{(1, 7)} = 0.39$, $p = 0.85$, ($1\text{-}\beta = 0.05$)
Task x Surface - $F_{(1, 7)} = 0.84$, $p = 0.39$, ($1\text{-}\beta = 0.13$)
Pace x Surface - $F_{(1, 7)} = 0.38$, $p = 0.851$, ($1\text{-}\beta = 0.05$)

Main effect for Task - $F_{(1, 7)} = 13.335$, $p = 0.008$*, ($1\text{-}\beta = 0.88$)

Post hoc – no significant differences (see table K.2)

One-way ANOVA for Pace – No significant effects
All comparisons - $F_{(1, 8)} \geq 0.02$, $p \geq 0.263$, ($1\text{-}\beta > 0.07$)

One– way ANOVA for Surface – No significant effects
All comparisons - $F_{(1, 8)} \geq 1.281$, $p \geq 0.05$, ($1\text{-}\beta > 0.05$)

Sol- Soleus* – $p < 0.05$, † - $p < 0.0125$, ($1\text{-}\beta$) – Observed Power, DL - Double-Leg hopping, AL – Alternate-leg hopping, Flat – Flat hopping surface, Dec – Decline hopping surface, S – Self-Selected pace, F – Forced pace

Figure K.2 Three-way ANOVA for left and right soleus for the derived variable of peak activation amplitude.

Table K.1 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of left soleus peak activation amplitude between flat and decline hopping surfaces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL S	0.01 (0.02)	0.01 [-0.01 to 0.02]	8	1.224	0.256	0.1
AL S	0.004 (0.01)	0.00 [0.00 to 0.01]	8	1.637	0.140	0.07
DL F	0.02 (0.03)	0.03 [0.00 to 0.04]	8	1.820	0.106	0.19
AL F	0.01 (0.01)	0.00 [0.00 to 0.02]	8	3.633	0.007*	0.47

*- $p < 0.125$, DL – Double-Leg, AL – Alternate-leg, S- Self-Selected Pace, F – Forced Pace

Table K.2 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of right soleus peak activation amplitude between double-leg and alternate-leg hopping tasks.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
Flat S	0.01 (0.03)	0.01 [-0.01 to 0.04]	8	1.405	0.198	0.05
Dec S	-0.00 (0.01)	0.01 [-0.01 to 0.01]	8	-0.560	0.593	0.02
Flat F	0.01 (0.01)	0.00 [0.00 to 0.02]	8	1.492	0.174	0.47
Dec F	0.00 (0.03)	0.01 [-0.02 to 0.02]	8	0.173	0.867	0.01

*- $p < 0.0125$, Dec – Decline, S- Self-Selected Pace, F – Forced Pace

Left MG 3-Way ANOVA - $F_{(1, 8)} = 2.466$, $p = 0.155$, $(1-\beta = 0.28)$

No 2-Way Interactions
All comparisons - $F_{(1, 8)} \geq 0.013$, $p \geq 0.194$, $(1-\beta > 0.05)$

No main effects
All comparisons - $F_{(1, 8)} \geq 0.172$, $p \geq 0.194$, $(1-\beta = 0.07)$

Right MG - 3-Way ANOVA - $F_{(1, 8)} = 2.187$, $p = 0.177$, $(1-\beta = 0.26)$

No 2-Way Interactions
All comparisons - $F_{(1, 8)} \geq 0.002$, $p \geq 0.139$, $(1-\beta > 0.05)$

No main effects
All comparisons - $F_{(1, 8)} \geq 2\ 2.124$, $p \geq 0.052$, $(1-\beta > 0.25)$

MG – medial gastrocnemius, * – $p < 0.05$, (1-β) – Observed Power, DL - Double-Leg hopping, AL – Alternate-leg hopping, Flat – Flat hopping surface, Dec – Decline hopping surface, S – Self-Selected pace, F – Forced pace

Figure K.3 Three-way ANOVA for left and right medial gastrocnemius (MG) for the derived variable of peak activation amplitude.

Left BF 3-Way ANOVA - $F_{(1, 8)} = 0.666$, $p = 0.438$, $(1\text{-}\beta = 0.11)$

No 2-Way Interactions

All comparisons - $F_{(1, 8)} \geq 0.106$, $p \geq 0.32$, $(1\text{-}\beta > 0.06)$

Main effect for Task - $F_{(1, 8)} = 8.206$, $p = 0.021$*, $(1\text{-}\beta = 0.71)$

Post hoc – No significant differences (see table K.3)

Right BF - 3-Way ANOVA - $F_{(1, 7)} = 0.277$, $p = 0.615$, $(1\text{-}\beta = 0.07)$

No 2-Way Interactions

All comparisons - $F_{(1, 7)} \geq 0.063$, $p \geq 0.148$, $(1\text{-}\beta > 0.06)$

No main effect

All comparisons - $F_{(1, 7)} \geq 0.144$, $p \geq 0.18$, $(1\text{-}\beta > 0.06)$

BF – biceps femoris, * – $p < 0.05$, $(1\text{-}\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – Flat hopping surface, Dec – Decline hopping surface, S – Self-Selected pace, F – Forced pace

Figure K.4 Three-way ANOVA for left and right biceps femoris for the derived variable of peak activation amplitude.

Table K.3 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of left biceps femoris peak activation amplitude between double-leg and alternate-leg hopping tasks.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
Flat S	-0.01 (0.03)	0.01 [-0.03 to 0.01]	8	-0.846	0.422	0.05
Dec S	-0.02 (0.03)	0.01 [-0.04 to 0.00]	8	-2.239	0.055	0.20
Flat F	-0.01 (0.02)	0.01 [-0.03 to 0.01]	8	-1.078	0.313	0.10
Dec F	-0.01 (0.05)	0.02 [-0.05 to 0.04]	8	-0.344	0.740	0.02

Dec – Decline, S- Self-Selected Pace, F – Forced Pace

Left VL 3-Way ANOVA - $F_{(1, 7)} = 0.268$, $p = 0.621$, $(1-\beta = 0.07)$

2-Way Interactions for Pace x Task - $F_{(1, 7)} = 9.717$, $p = 0.017$*, $(1-\beta = 0.76)$

Main effect for surface - $F_{(1, 7)} = 12.753$, $p = 0.009$*, $(1-\beta = 0.86)$

Post hoc – AL at S (see table K.4)

One-way ANOVA for Pace
AL on Flat - $F_{(1, 8)} = 28.102$, $p = 0.001$*, $(1-\beta = 0.99)$
AL on Decline - $F_{(1, 7)} = 14.66$, $p = 0.006$*, $(1-\beta = 0.91)$

One-way ANOVA for Task
Flat at S – $F_{(1, 8)} = 18.669$, $p = 0.003$*, $(1-\beta = 0.96)$
Decline at S - $F_{(1, 8)} = 6.984$, $p = 0.03$*, $(1-\beta = 0.64)$

Right VL 3-Way ANOVA - $F_{(1, 8)} = 1.197$, $p = 0.306$, $(1-\beta = 0.16)$

2-Way Interactions for Pace x Task - $F_{(1, 8)} = 18.011$, $p = 0.003$, $(1-\beta = 0.96)$

No main effects
All comparisons - $F_{(1, 8)} \geq 0.425$, $p \geq 0.014$, $(1-\beta > 0.09)$

One-way ANOVA for Pace
AL Flat - $F_{(1, 8)} = 21.968$, $p = 0.002$*, $(1-\beta = 0.98)$
AL Decline - $F_{(1, 8)} = 11.769$, $p = 0.009$*, $(1-\beta = 0.85$

One-way ANOVA for Task
Flat at S - $F_{(1, 8)} = 8.52$, $p = 0.019$*, $(1-\beta = 0.73)$
Dec at S - $F_{(1, 8)} = 8.038$, $p = 0.022$*, $(1-\beta = 0.70)$

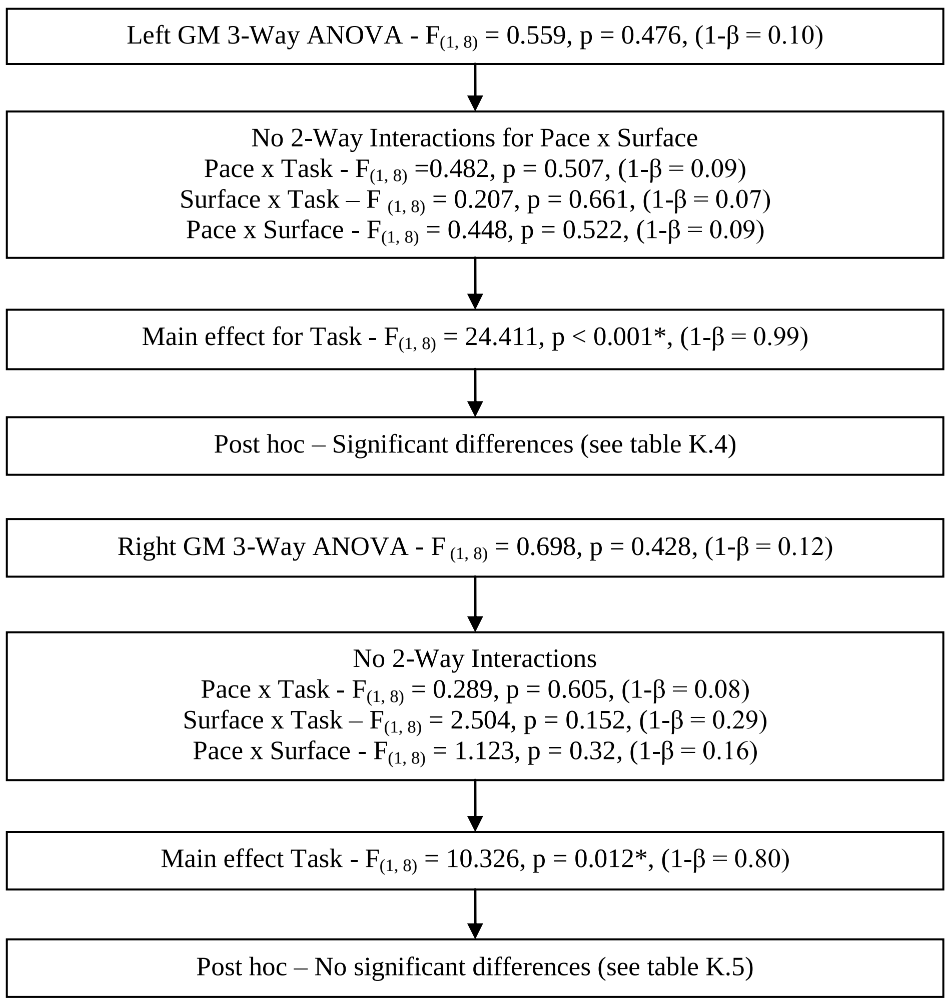
VL – vastus lateralis, * – $p < 0.05$, $(1-\beta)$ –Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – Flat hopping surface, Dec – Decline hopping surface,
S – Self-Selected pace, F – Forced pace

Figure K.5 Three-way ANOVA for left and right vastus lateralis for the derived variable of peak activation amplitude.

Table K.4 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of left vastus lateralis peak activation amplitude between flat and decline hopping surfaces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL S	-0.01 (0.04)	0.01 [-0.04 to 0.03]	8	-0.382	0.712	0.03
AL S	-0.03 (0.02)	0.01 [-0.05 to -0.01]	8	-3.825	0.005*	0.86
DL F	-0.01 (0.03)	0.01 [-0.04 to 0.02]	7	-0.974	0.362	0.05
AL F	-0.02 (0.03)	0.01 [-0.04 to 0.00]	7	-2.482	0.042	0.19

*- p < 0.0125, S- Self-Selected Pace, F – Forced Pace, DL – Double-Leg, AL – Alternate-Leg



GM – gluteus medius, * – p < 0.05, (1-β) –Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – Flat hopping surface, Dec – Decline hopping surface, S – Self-Selected pace, F – Forced pace

Figure K.6 Three-way ANOVA for left and right gluteus medius (GM) for the derived variable of peak activation amplitude.

Table K.4 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of left gluteus medius peak activation amplitude between double-leg and alternate-leg hopping tasks.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
Flat S	-0.04 (0.05)	0.02 [-0.07 to 0.00]	8	-2.224	0.057	0.29
Dec S	-0.05 (0.03)	0.01 [-0.07 to -0.03]	8	-5.006	0.001*	0.93
Flat F	-0.06 (0.05)	0.02 [-0.10 to -0.02]	8	-3.331	0.010*	0.65
Dec F	-0.05 (0.06)	0.02 [-0.10 to -0.01]	8	-2.672	0.028	0.32

*- $p < 0.0125$, S- Self-Selected Pace, F – Forced Pace, Dec - Decline

Table K.5 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of right gluteus medius peak activation amplitude between double-leg and alternate-leg hopping tasks.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
Flat S	-0.02 (0.06)	0.02 [-0.07 to 0.03]	8	-1.028	0.334	0.05
Dec S	-0.05 (0.05)	0.02 [-0.08 to -0.01]	8	-2.908	0.020	0.47
Flat F	-0.04 (0.06)	0.02 [-0.09 to 0.00]	8	-2.111	0.068	0.19
Dec F	-0.06 (0.08)	0.03 [-0.12 to 0.00]	8	-2.160	0.063	0.25

*- $p < 0.0125$, S- Self-Selected Pace, F – Forced Pace, Dec - Decline

Left MG 3-Way ANOVA - $F_{1, 8} = 0.901$, $p = 0.37$, $(1-\beta = 0.13)$

↓

2-Way Interactions
Pace x Task - $F_{1, 8} = 3.958$, $p = 0.082$, $(1-\beta = 0.42)$
Task x Surface - $F_{1, 8} = 0.184$, $p = 0.679$, $(1-\beta = 0.07)$
Pace x Surface - $F_{1, 8} = 0.015$, $p = 0.906$, $(1-\beta = 0.05)$

↓

Main effect surface - $F_{1, 8} = 21.866$, $p = 0.002$*, $(1-\beta = 0.98)$

↓

Post hoc – significant differences (see table K.6)

Right MG 3-Way ANOVA - $F_{1, 8} = 1.45$, $p = 0.263$, $(1-\beta = 0.19)$

↓

No 2-Way Interactions
Pace x Surface – $F_{1, 8} = 0.035$, $p = 0.857$, $(1-\beta = 0.05)$
Task x Surface – $F_{1, 8} = 0.898$, $p = 0.371$, $(1-\beta = 0.13)$
Pace x Task – $F_{1, 8} = 1.111$, $p = 0.0323$, $(1-\beta = 0.15)$

↓

Main effect for hopping surface - $F_{1, 8} = 6.679$, $p = 0.032$*, $(1-\beta = 0.62)$

↓

Post hoc – no significant differences (see table K.7)

MG – medial gastrocnemius, * – $p < 0.05$, $(1-\beta)$ –Observed Power, DL - Double-Leg hopping, AL – Alternate-leg hopping, Flat – Flat hopping surface, Dec – Decline hopping surface, S – Self-Selected pace, F – Forced pace

Figure K.7 Three-way ANOVA for left and right medial gastrocnemius for the derived variable of normalised pre-loading activation amplitude.

Table K.6 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of left medial gastrocnemius normalised pre-loading activation amplitude between flat and decline hopping surfaces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL S	-15.15 (11.65)	3.88 [-24.11 to -6.19]	8	-3.899	0.005*	0.74
AL S	-14.62 (12.98)	4.33 [-24.60 to -4.65]	8	-3.381	0.010*	0.59
DL F	-12.12 (17.20)	5.73 [-25.34 to 1.09]	8	-2.115	0.067	0.22
AL F	-16.60 (13.13)	4.38 [-26.69 to -6.50]	8	-3.791	0.005*	0.71

* - $p < 0.0125$, DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace, F – Forced Pace

Table K.7 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of right medial gastrocnemius normalised pre-loading activation amplitude between flat and decline hopping surfaces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
DL S	-18.55 (19.65)	6.55 [-33.65 to -3.45]	8	-2.832	0.022	0.42
AL S	-9.68 (25.11)	8.37 [-28.99 to -9.62]	8	-1.157	0.281	0.06
DL F	-14.48 (20.91)	6.97 [-30.55 to 1.59]	8	-2.077	0.071	0.21
AL F	-15.81 (16.71)	5.57 [-28.65 to -2.97]	8	-2.840	0.022	0.42

* - $p < 0.0125$, DL – Double-Leg, AL – Alternate-Leg, S – Self-Selected Pace, F – Forced Pace

Left CR_a 1-Way ANOVA – $F_{(6, 48),}$ 48 = 3.069, $p = 0.013$*, $(1-\beta = 0.87)$

No significant pairwise comparisons ($p > 0.05$)
Multivariate tests – $F_{(6, 3)} = 5.047$, $p = 1.06$, $(1-\beta = 0.42)$

Right CR_a 1-Way ANOVA – $F_{(6, 42)} = 2.326$, $p = 0.157$, $(1-\beta = 0.32)$

CR_a - Co-activation ratio between soleus & tibialis anterior,
* – $p < 0.05$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping,
Flat – Flat hopping surface, Dec – Decline hopping surface, S – Self-Selected pace, F – Forced pace

Figure K.8 One-way ANOVA for left co-activation ratio between soleus and tibialis anterior.

Left CR_b 1-Way ANOVA – $F_{(6, 48)} = 2.884$, $p = 0.057$, $(1-\beta = 0.61)$

Right CR_b 1-Way ANOVA - $F_{(6, 48)} = 2.349$, $p = 0.137$, $(1-\beta = 0.37)$

CR_b – Co-activation Ratio between medial gastrocnemius and tibialis anterior,
$(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping, Flat – Flat hopping surface, Dec – Decline hopping surface, S – Self-Selected pace, F – Forced pace

Figure K.9 One-way ANOVA for left and right co-activation ratio between medial gastrocnemius and tibialis anterior.

Left CR_c 1-Way ANOVA - $F_{(6, 42)} = 3.101$, $p = 0.068$, $(1-\beta = 0.54)$

Right CR_c 1-Way ANOVA - $F_{(6, 42)} = 5.912$, $p = 0.024$*, $(1-\beta = 0.70)$

No significant pairwise comparisons ($p \geq 0.417$)
Multivariate tests – $F_{(6, 2)} = 0.877$, $p = 0.619$, $(1-\beta = 0.09)$

CR_c – Co-activation Ratio between vastus lateralis and biceps femoris,
*– $p < 0.05$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping,, Flat – Flat hopping surface, Dec – Decline hopping surface, S – Self-Selected pace, F – Forced pace

Figure K.10 One-way ANOVA for left and right co-activation ratio between vastus lateralis and biceps femoris

Left CR_d 1-Way ANOVA - $F_{(6, 42)} = 6.403$, $p = 0.021$*, $(1-\beta = 0.72)$

↓

Significant pairwise comparisons (see table K.8)

Right CR_d 1-Way ANOVA - $F_{(6, 48)} = 7.431$, $p = 0.003$*, $(1-\beta = 0.92)$

↓

Significant pairwise comparisons (see table K.9)

CR_d – Co-activation Ratio between vastus lateralis and medial gastrocnemius,
*– $p < 0.05$, $(1-\beta)$ – Observed Power, DL - Double-Leg hopping, AL –Alternate-leg hopping,, Flat – Flat hopping surface, Dec – Decline hopping surface, S – Self-Selected pace, F – Forced pace

Figure K.11 One-way ANOVA for left and right co-activation ratio between vastus lateralis and medial gastrocnemius.

Table K.8 Results of pairwise comparisons (mean difference (SD), SEM [95%CI], p value and observed power (1-β)) of the left co-activation ratio (vastus lateralis and medial gastrocnemius) between self-selected and forced hopping paces for alternate-leg hopping on flat and decline surfaces.

Hopping condition	Mean Difference (SD)	Standard error of the mean [95%CI]	p	1-β
AL Flat S – AL Flat F	-0.67 (0.32)	0.114 [-1.2 to -0.14]	0.013*	1.0
AL Dec S – AL Dec F	-0.62 (0.34)	-1.169 [-28.99 to -0.064]	0.027*	1.0
AL Flat S – AL Dec F	-0.933 (0.58)	0.119 [-1.183 to -.0383]	0.002*	0.99

* - $p < 0.05$, AL – Alternate-Leg, Dec – Decline, S – Self-Selected Pace, F – Forced Pace

Table K.9 Results of pairwise comparisons (mean difference (SD), SEM [95%CI], p value and observed power (1-β)) of the right co-activation ratio (vastus lateralis and medial gastrocnemius) between self-selected and forced hopping paces for double-leg hopping on flat and decline surfaces.

Hopping conditions	Mean Difference (SD)	Standard error of the mean [95%CI]	p	1-β
DL Dec S – AL Flat S	0.57 (0.32)	0.107 [0.102 to 1.038]	0.015*	1.0
DL Dec S – DL Flat F	-0.07 (0.41)	0.14 [-0.67 to 0.53]	1.0	0.07
DL Dec S – DL Dec F	-0.43 (0.38)	0.13 [-0.98 to 0.13]	0.206	0.98
DL Dec S – AL Dec F	-0.30 (0.46)	0.16 [-0.97 to 0.38]	1.0	0.41

* - $p < 0.05$, DL – Double-Leg, AL – Alternate-Leg, Dec – Decline, S- Self-Selected Pace, F – Forced Pace

Appendix L - Calibration of KinCom Dynamometer prior to data collection

Left side: $y = 17.849x + 25.352$ ($R^2 = 1.0$)

Right side: $y = -17.267x - 23.946$ ($R^2 = 1.0$)

Where y is in kg, x is in volts.

Appendix M - Victorian Institute of Sport Assessment (VISA)

1. For how many minutes can you sit pain-free? ☐

Points ☐☐☐☐☐☐☐☐☐☐☐

0 mins — 100 mins

0 1 2 3 4 5 6 7 8 9 10

2. Do you have pain walking downstairs with a normal gait cycle? ☐

Points ☐☐☐☐☐☐☐☐☐☐☐

Strong severe pain — No pain

0 1 2 3 4 5 6 7 8 9 10

3. Do you have pain at the knee with full active non-weight bearing knee extension? ☐

Points ☐☐☐☐☐☐☐☐☐☐☐

Strong severe pain — No pain

0 1 2 3 4 5 6 7 8 9 10

4. Do you have pain when doing a full weight-bearing lunge? ☐

Points ☐☐☐☐☐☐☐☐☐☐☐

Strong severe pain — No pain

0 1 2 3 4 5 6 7 8 9 10

5. Do you have problems squatting? ☐

Points ☐☐☐☐☐☐☐☐☐☐☐

Unable — No problem

0 1 2 3 4 5 6 7 8 9 10

6. Do you have pain during or immediately after doing 10 single leg hops? ☐

Points ☐☐☐☐☐☐☐☐☐☐☐

Strong severe pain/ unable — No pain

0 1 2 3 4 5 6 7 8 9 10

7. Are you currently undertaking sport or other physical activity? Points ☐

0 ☐ Not at all
4 ☐ Modified training ± modified competition
7 ☐ Full training ± competition but not at the same level as when symptoms began
10 ☐ Competing at the same or higher level when symptoms began

8. Please complete EITHER A, B or C in this question.

- If you have **no pain** while undertaking sport please complete **Q8a only**
- If you have **pain while undertaking sport but it does not stop you** from completing the activity, please complete **Q8b only.**
- If you **have pain that stops you from completing sporting activities,** please complete Q**8 c only**

8a. If you have no pain while undertaking sport, for how long can you train/practise?

NIL	0-5 mins	5-10 mins	11-15 mins	>15 mins
☐	☐	☐	☐	☐
0	**7**	**14**	**21**	**30**

Points ☐

or

8b. If you have some pain while undertaking sport, but it does not stop you from completing your training/practice, for how long can you train/practise?

NIL	0-5 mins	5-10 mins	11-15 mins	>15 mins
☐	☐	☐	☐	☐
0	**4**	**10**	**14**	**20**

Points ☐

or

8c. If you have pain that stops you from completing your training/practice, for how long can you train/practise?

NIL	0-5 mins	5-10 mins	11-15 mins	>15 mins
☐	☐	☐	☐	☐
0	**2**	**5**	**7**	**10**

Points ☐

TOTAL VISA SCORE ☐

Appendix N - Statistical results for comparison of patellar tendon mechanical properties (chapter 6)

2-Way ANOVA – $F_{(3, 47)} = 0.393$, $p = 0.759$, ($1-\beta = 0.12$)

↓

Main effect
Side - $F_{(1, 3)} = 7.01$, $p = 0.011$*, ($1-\beta = 0.74$) (see Table N.1)
Group - $F_{(3, 47)} = 1.59$, $p = 0.204$, ($1-\beta = 0.39$)

*- $p < 0.05$, ($1-\beta$) – Observed Power

Figure N.1 Two-way ANOVA for normalised patellar tendon force.

Table N.1 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of normalised patellar tendon force ($N.kg^{-2/3}$) between left and right sides for each sporting group.

Group	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
♂ VB	-13.68 (46.75)	11.02 [-36.93 to 9. 57]	17	-1.242	0.231	0.07
♂ BB	-7.40 (32.86)	9.11 [-27.26 to 12.45]	12	-0.812	0.432	0.03
♀ BB	-12.93 (35.12)	9.39 [-33.21 to 7.35]	13	-1.377	0.192	0.09
SW	-27.96 (25.92)	10.58 [-55.16 to -0.76]	5	-2.643	0.046	0.23

VB –Volleyball, BB –Basketball, SW- Swimming

2-Way ANOVA – $F_{(3, 47)} = 0.179$, $p = 0.91$, ($1-\beta = 0.08$)

↓

Main effect
Side - $F_{(1, 3)} = 0.053$, $p = 0.819$, ($1-\beta = 0.06$)
Group - $F_{(3, 47)} = 0.179$, $p = 0.91$, ($1-\beta = 0.66$)

*- $p < 0.05$, ($1-\beta$) – Observed Power

Figure N.2 Two-way ANOVA for patellar tendon proximal cross-sectional area.

2-Way ANOVA – $F_{(3, 47)} = 1.098$, $p = 0.359$, ($1-\beta = 0.06$)

↓

Main effect
Side - $F_{(1, 3)} = 0.078$, $p = 0.781$, ($1-\beta = 0.28$)
Group - $F_{(3, 47)} = 2.28$, $p = 0.092$, ($1-\beta = 0.54$)

*- $p < 0.05$, ($1-\beta$) – Observed Power

Figure N.3 Two-way ANOVA for patellar tendon strain.

Table N.2 Two-way ANOVA ($F_{(\text{degrees of freedom})}$ test, p value and observed power (1-β)) for comparison of patellar tendon stiffness (10% intervals of maximal voluntary isometric contraction (MVIC)) between sides (2) and groups (4).

10%	F test		
interval	Group by Side	Main effect (side)	Main effect (group)
20 to 30	$F_{(3, 47)} = 0.29$, $p = 0.83_{1-\beta = 0.10}$	$F_{(1, 47)} = 0.659$, $p = 0.42_{1-\beta = 0.13}$	$F_{(3, 47)} = 1.571$, $p = 0.21_{1-\beta = 0.39}$
30 to 40	$F_{(3, 47)} = 2.158$, $p = 0.10_{1-\beta = 0.52}$	$F_{(1, 47)} = 0.287$, $p = 0.60_{1-\beta = 008}$	$F_{(3, 47)} = 0.794$, $p = 0.5_{1-\beta = 0.21}$
40 to 50	$F_{(3, 47)} = 1.58$, $p = 0.21_{1-\beta = 0.39}$	$F_{(1, 47)} = 0.292$, $p = 0.59_{1-\beta = 0.08}$	$F_{(3, 47)} = 0.167$, $p = 0.92_{1-\beta = 0.08}$
50 to 60	$F_{(3, 47)} = 0.949$, $p = 0.42_{1-\beta = 0.24}$	$F_{(1, 47)} = 0.257$, $p = 0.62_{1-\beta = 0.08}$	$F_{(3, 47)} = 0.190$, $p = 0.90_{1-\beta = 0.08}$
60 to 70	$F_{(3, 47)} = 0.661$, $p = 0.58_{1-\beta = 0.18}$	$F_{(1, 47)} = 1.362$, $p = 0.25_{1-\beta = 0.21}$	$F_{(3, 47)} = 0.766$, $p = 0.52_{1-\beta = 0.20}$
70 to 80	$F_{(3, 47)} = 0.984$, $p = 0.41_{1-\beta = 0.25}$	$F_{(1, 47)} = 0.361$, $p = 0.55_{1-\beta = 0.09}$	$F_{(3, 47)} = 1.272$, $p = 0.30_{1-\beta = 0.32}$

2-Way ANOVA – $F_{(3, 47)} = 0.581$, $p = 0.581$, $(1-\beta = 0.18)$

↓

Main effects
Group - $F_{(3, 47)} = 4.616$, $p = 0.007$*, $(1-\beta = 0.64)$ (see table N.3)
Side - $F_{(1, 47)} = 5.559$, $p = 0.023$*, $(1-\beta = 0.86)$ (see table N.4)

*- $p < 0.05$, (1-β) – Observed Power

Figure N.4 Two-way ANOVA for patellar tendon stress.

Table N.3 Pairwise comparisons (mean difference (MPa), SEM [95%CI, p value and observed power (1-β)) for between group differences for comparison of patellar tendon stress (MPa).

Comparison	Mean Difference	Standard error of the mean [95%CI]	p	1-β
♂VB v ♂BB	6.73	2.44 [0.021 to 13.44]	0.049*	0.64
♂VB v ♀BB	7.20	2.34 [0.63 to 133.77]	0.025*	0.83

*- $p < 0.05$, VB – Volleyball, BB –Basketball, SW- Swimming

Table N.4 Post hoc paired t-test results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of patellar tendon stress (MPa) between left and right sides for each sporting group.

Group	Mean Difference (SD)	Standard error of the mean [95%CI]	df	t	p	1-β
♂ VB	-1.71 (7.07)	1.67 [-5.23 to 1.81]	17	-1.027	0.32	0.05
♂ BB	0.58 (3.93)	1.09 [-2.996 to 1.80]	12	-0.531	0.61	0.02
♀ BB	-1.09 (3.67)	0.98 [-3.21 to 1.03]	13	-1.113	0.29	0.06
SW	-4.05 (3.09)	1.26 [-7.29 to -0.80]	5	-2.643	0.02	0.36

VB – Volleyball, BB –Basketball, SW- Swimming

Table N.5 Two-way ANOVA ($F_{(degrees\ of\ freedom)}$ test, p value and observed power (1-β)) for comparison of patellar tendon elastic modulus (10% intervals of maximal voluntary isometric contraction (MVIC)) between sides (2) and groups (4).

10% interval	F test		
	Group by Side	Main effect (side)	Main Effect (group)
20 to 30	$F_{(3, 47)} = 0.457$, $p = 0.71_{1-\beta = 0.14}$	$F_{(1, 47)} = 0.328$, $p = 0.57_{1-\beta = 0.09}$	$F_{(3, 47)} = 1.798$, $p = 0.16_{1-\beta = 0.44}$
30 to 40	$F_{(3, 47)} = 2.792$, $p = 0.05_{1-\beta = 0.64}$	$F_{(1, 47)} = 0.292$, $p = 0.59_{1-\beta = 0.08}$	$F_{(3, 47)} = 1.080$, $p = 0.37_{1-\beta = 0.27}$
40 to 50	$F_{(3, 47)} = 2.180$, $p = 0.10_{1-\beta = 0.52}$	$F_{(1, 47)} = 0.162$, $p = 0.70_{1-\beta = 0.07}$	$F_{(3, 47)} = 0.352$, $p = 0.79_{1-\beta = 0.11}$
50 to 60	$F_{(3, 47)} = 2.294$, $p = 0.09_{1-\beta = 0.54}$	$F_{(1, 47)} = 0.110$, $p = 0.74_{1-\beta =0.06}$	$F_{(3, 47)} = 0.371$, $p = 0.77_{1-\beta = 0.12}$
60 to 70	$F_{(3, 47)} = 0.839$, $p = 0.48_{1-\beta = 0.22}$	$F_{(1, 47)} = 0.198$, $p = 0.66_{1-\beta = 0.07}$	$F_{(3, 47)} = 0.314$, $p = 0.82_{1-\beta = 0.11}$
70 to 80	$F_{(3, 47)} = 1.343$, $p = 0.27_{1-\beta = 0.33}$	$F_{(1, 47)} = 0.010$, $p = 0.92_{1-\beta = 0.05}$	$F_{(3, 47)} = 0.715$, $p = 0.55_{1-\beta = 0.19}$

Table N.6 Multiple independant t-tests ($t_{(degrees\ of\ freedom)}$ test, mean difference, p value, SEM [95%CI] and observed power (1-β)) for comparison of men's volleyball players that prefer to land on their left leg to those that prefer to land on their right leg for left and right patellar tendon stiffness ($N.mm^{-1}$) (10% intervals of MVIC).

10% interval	t-test	Mean Difference	Standard Error [95%CI]	1-β
L 20 to 30	$t_{(16)} = 0.994$, $p =0.335$	113.81	114.44 [-128.80 to 356.42]	0.15
L 30 to 40	$t_{(16)} = 0.704$, $p = 0.491$	142.69	202.60 [286.81 to 572.19]	0.11
L 40 to 50	$t_{(16)} = 0.398$, $p = 0.696$	84.81	213.25 [-367.26 to 536.88]	0.07
L 50 to 60	$t_{(16)} = -0.112$, $p = 0.912$	-32.92	293.46 [-655.02 to 589.19]	0.05
L 60 to 70	$t_{(16)} = -0.742$, $p = 0.469$	-362.12	487.94 [-1396.51 to 672.27]	0.09
L 70 to 80	$t_{(16)} = -2.285$, $p = 0.036$*	-1326.27	580.45 [-2556.78 to -95.76]	0.51
R 20 to 30	$t_{(16)} = -0.284$, $p = 0.78$	-51.54	181.67 [-436.63 to 333.55]	0.06
R 30 to 40	$t_{(16)} = -0.141$, $p = 0.889$	-21.15	206.05 [-465.97 to 407.66]	0.05
R 40 to 50	$t_{(16)} = -0.325$, $p = 0.749$	-85.45	262.86 [-642.87 to 471.78]	0.06
R 50 to 60	$t_{(16)} = -1.01$, $p = 0.328$	-357.43	354.04 [-1107.96 to 393.11]	0.17
R 60 to 70	$t_{(16)} = 0.773$, $p = 0.451$	416.80	439.10 [-726.04 to 1559.64]	0.12
R 70 to 80	$t_{(16)} = 0.321$, $p = 0.753$	262.12	817.70 [-1471.32 to 1995.46]	0.06

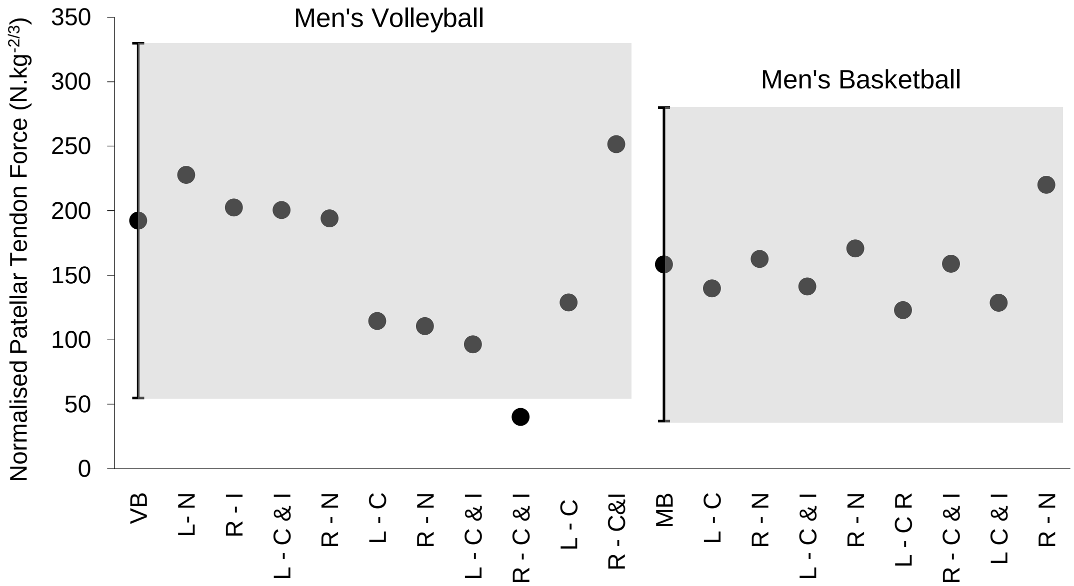
*- $p < 0.05$, L – Left, R – Right

Table N.7 Multiple independant t-tests ($t_{(degrees\ of\ freedom)}$ test, mean difference, p value, SEM [95%CI] and observed power (1-β)) for comparison of men's volleyball players that prefer to land on their left leg to those that prefer to land on their right leg for left and right patellar tendon elastic modulus (GPa) (10% intervals of MVIC).

10% interval	t-test	Mean Difference	Standard Error [95%CI]	1-β
L 20 to 30	$t_{(16)} = 0.159$, $p =0.876$	0.01	0.04 [-0.07 to 0.09]	0.05
L 30 to 40	$t_{(16)} = 0.095$, $p = 0.926$	0.01	0.05 [-0.11 to 0.12]	0.05
L 40 to 50	$t_{(16)} = -0.01$, $p = 0.992$	-0.001	0.08 [-0.18 to 0.18]	0.05
L 50 to 60	$t_{(16)} = -0.512$, $p = 0.616$	-0.05	0.1 [-0.26 to 0.16]	0.08
L 60 to 70	$t_{(16)} = -0.948$, $p = 0.357$	-0.17	0.18 [-0.55 to 0.21]	0.12
L 70 to 80	$t_{(16)} = -2.1$, $p = 0.052$	-0.45	0.21 [-0.91 to 0.004]	0.56
R 20 to 30	$t_{(16)} = -0.154$, $p = 0.88$	-0.01	0.06 [-0.13 to 0.11]	0.05
R 30 to 40	$t_{(16)} = -0.118$, $p = 0.907$	-0.01	0.06 [-0.13 to 0.12]	0.05
R 40 to 50	$t_{(16)} = -0.22$, $p = 0.829$	-0.02	0.07 [-0.17 to 0.14]	0.21
R 50 to 60	$t_{(16)} = -1.009$, $p = 0.328$	-0.10	0.10 [-0.31 to 0.11]	0.13
R 60 to 70	$t_{(16)} = 0.619$, $p = 0.545$	0.08	0.13 [-0.19 to 0.35]	0.09
R 70 to 80	$t_{(16)} = 0.22$, $p = 0.829$	0.06	0.28 [-0.53to 0.56]	0.06

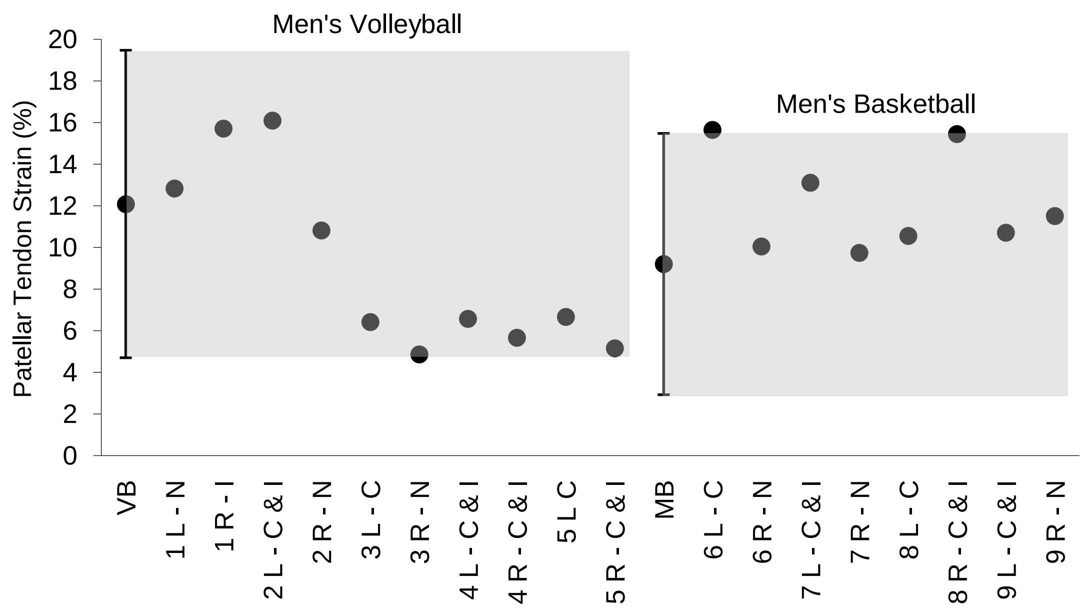
*- $p < 0.05$, L – Left, R – Right

Figure N.5a



L - Left   R - Right   N - Normal   C - Clinical Tendinopathy   I - Imaging Tendinopathy

Figure N.5b



L- Left   R - Right   C - Clinical Tendinopathy   I - Imaging Tendinopathy

Figure N.5c

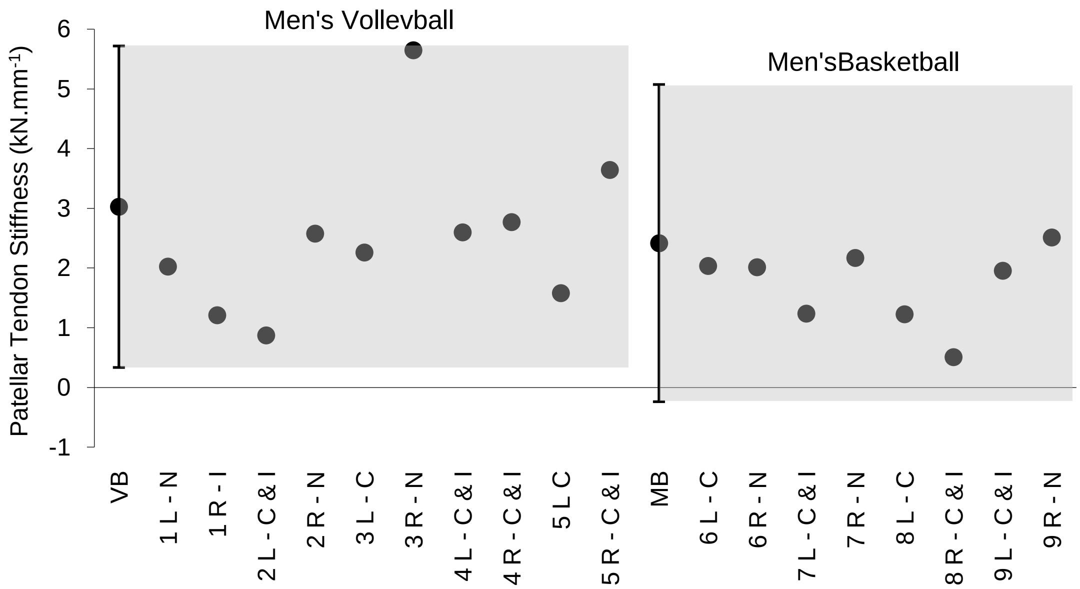


Figure N.5d

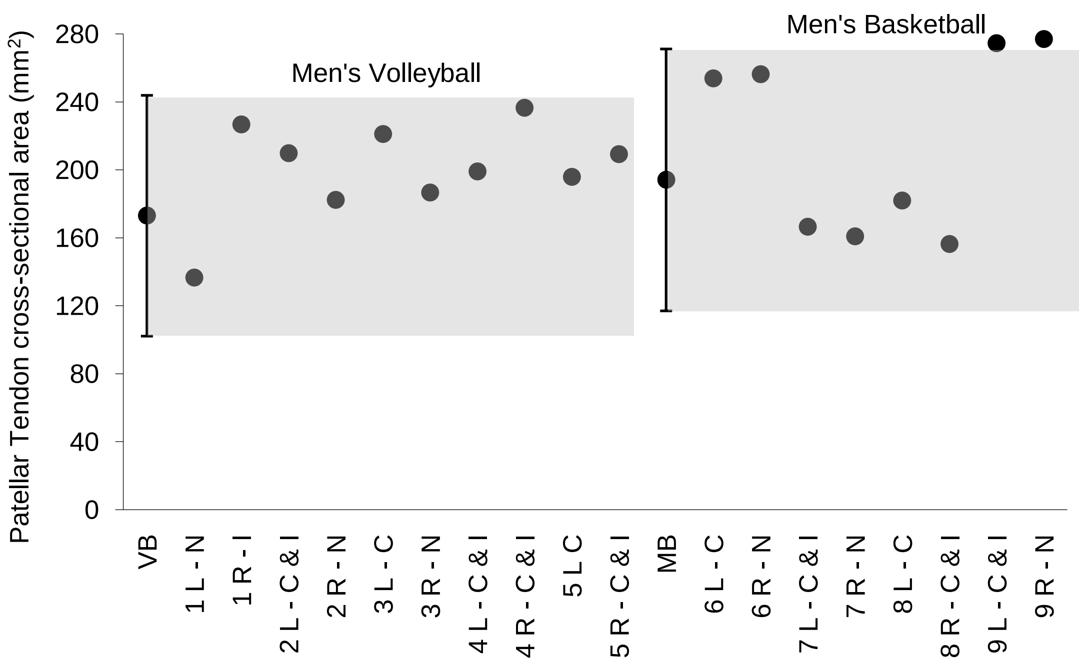


Figure N.5e

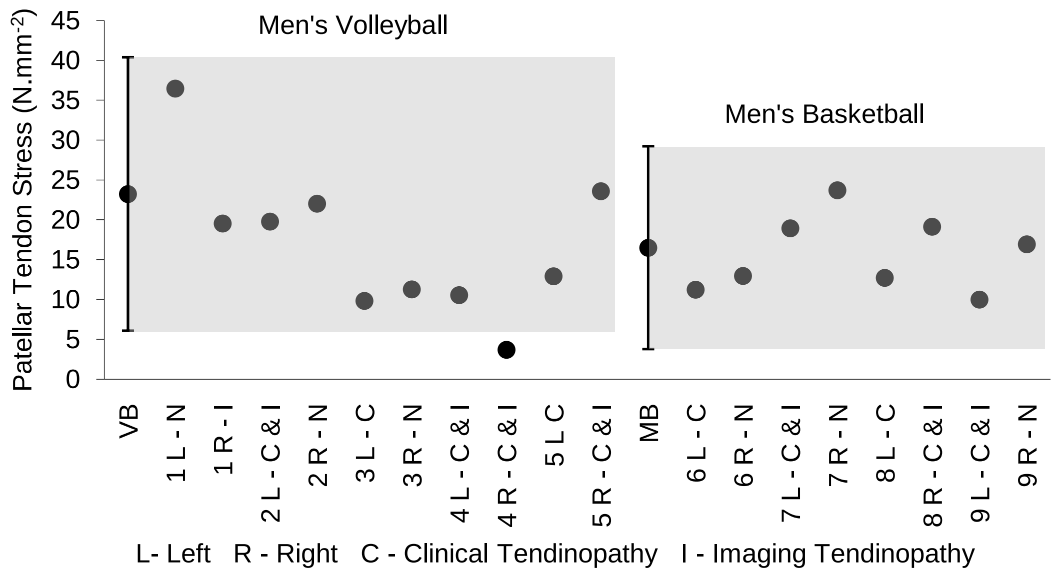


Figure N.5f

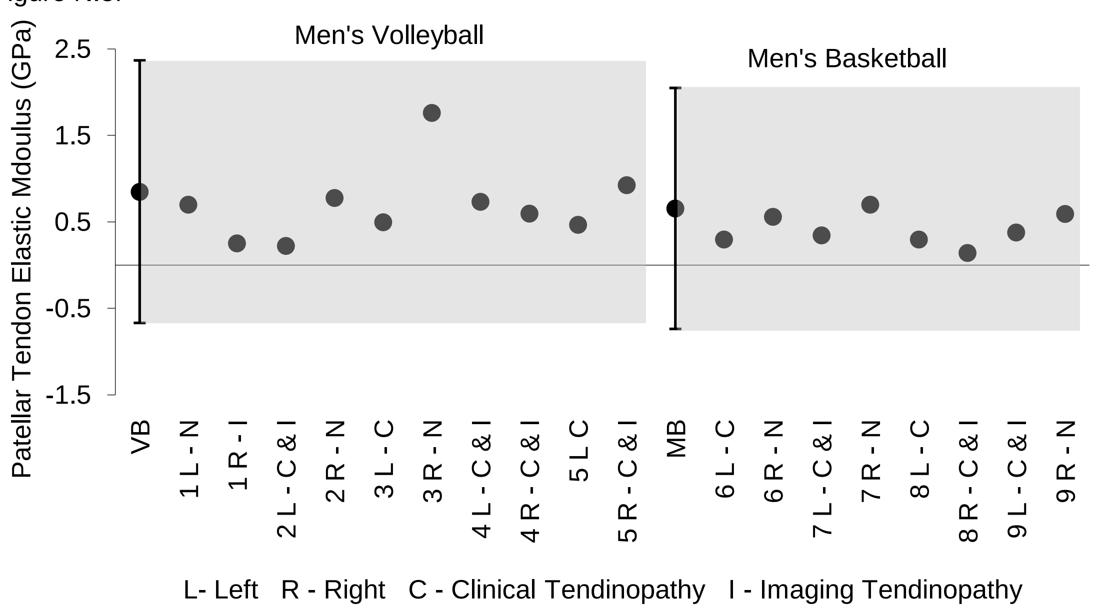


Figure N.5 Comparison of patellar tendon mechanical properties (normalised force (a), strain (b), stiffness (c), proximal cross-sectional area (d), stress (e) and elastic modulus (f)) for patellar tendinopathy (C- clinical tendinopathy, I – imaging tendinopathy and N – normal) against the mean and 95% CI for groups of men's volleyball and basketball players with normal patellar tendons bilaterally (shaded grey area).

Table N.8 Multiple paired t-tests (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) for comparison of patellar tendon cross-sectional area (mm^2) between left and right sides.

Group	Mean Difference (SD)	Standard error of the mean [95% CI]	df	t	p	1-β
♂VB_P	-2.42 (31.63)	7.46 [-18.15 to 13.31]	17	-0.325	0.749	0.06
♂BB_P	-6.45 (38.60)	10.71 [-29.77 to 16.88]	12	-0.602	0.558	0.09
♀BB_P	0.79 (30.84)	8.24 [-17.02 to 18.59]	13	0.095	0.926	0.05
SW_P	3.58 (16.29)	6.65 [-13.51 to 20.68]	5	-0.539	0.613	0.0.07
♂VB_M	1.98 (28.44)	6.94 [-12.66 to 16.61]	17	0.285	0.779	0.06
♂BB_M	-9.38 (29.05)	8.06 [-26.93 to 8.18]	12	-1.164	0.267	0.19
♀BB_M	2.65 (26.73)	7.14 [-12.78 to 18.08]	13	0.371	0.717	0.06
SW_M	-1.13 (11.17)	4.56 [-12.85 to 10.59]	5	-0.249	0.814	0.05
♂VB_D	-2.58 (35.13)	8.28 [-20.05 to 14.89]	17	-0.312	0. 759	0.06
♂BB_D	-14.92 (30.02)	8.32 [-33.06 to 3.23]	12	-1.791	0.099	0.06
♀BB_D	11.28 (43.25)	11.26 [-13.70 to 36.25]	13	0.976	0.347	0.15
SW_D	1.73 (10.26)	4.19 [-9.03 to 12.50]	5	-0.414	0.696	0.06

VB –Volleyball, BB –Basketball, SW – Swimming, $_P$ – Proximal, $_M$ – Mid, $_D$ – Distal

Table N.9 One-way ANOVA ($F_{(\text{degrees of freedom})}$ test, p value and observed power (1-β)) comparing patellar tendon cross-sectional area (mm^2) between three sites (proximal, mid and distal) for each sporting group.

Group	F test	1-β
Men's Volleyball	$F_{(2, 34)} = 3.014$, $p = 0.062$	0.55
Men's Basketball	$F_{(2, 24)} = 2.937$, $p = 0.072$	0.52
Women's Basketball	$F_{(2, 26)} = 2.242$, $p = 0.126$	0.42
Swimmers	$F_{(2, 10)} = 7.825$, $p = 0.009$* (see Table N.10)	0.87

*- $p < 0.05$

Table N.10 Pairwise comparisons (mean difference, SEM [95%CI], p value and observed power (1-β)) for between group differences for comparison of patellar tendon cross-sectional area (mm^2) across the three measurement sites (proximal, mid and distal) for the swimming group.

Comparison	Mean Difference (SD)	Standard error of the mean [CI]	p	1-β
Proximal v Mid	8.51 (3.92)	1.6 [2.855 to 14.162]	*0.009	0.99
Proximal v Distal	7.86 (7.97)	3.252 [-3.634 to 19.35]	0.181	0.50
Mid v Distal	-0.65 (4.94)	2.02 [-7.77 to 6.47]	1.00	0.06

*- $p < 0.05$

Appendix O - Statistical results for comparison of patellar tendon mechanical properties (chapter 7)

Table O.1 Paired t-tests results (mean difference (SD), SEM [95%CI], degrees of freedom (df), t value, p value and observed power (1-β)) comparing height and weight between sessions 1 and 2 for the left and right sides.

	Mean Difference (SD)	Standard error of the mean [CI]	df	t	p	1-β
Height (cm)	-1.638 (2.34)	0.65 [-3.09 to -0.26]	12	-2.586	0.024*	0.64
Weight (kg)	-1.79 (3.67)	1.02 [-4.01 to -0.43]	12	-1.759	0.104	0.37

*- $p < 0.05$

2x2 Way ANOVA – $F_{(1, 8)} = 0.005$, $p = 0.947$, $(1-\beta = 0.05)$

↓

Main effects
Side - $F_{(1, 8)} = 0.125$, $p = 0.732$, $(1-\beta = 0.06)$
Session - $F_{(1, 8)} = 0.626$, $p = 0.452$, $(1-\beta = 0.11)$

*- $p < 0.05$, $(1-\beta)$ – Observed Power

Figure O.1 Two by two ANOVA for normalised patellar tendon force (n=9).

2x2 Way ANOVA – $F_{(1, 8)} = 0.045$, $p = 0.837$, $(1-\beta = 0.05)$

↓

Main effects
Side - $F_{(1, 8)} > 1.252$, $p = 0.296$, $(1-\beta = 0.17)$
Session - $F_{(1, 8)} = 2.702$, $p = 0.139$, $(1-\beta = 0.31)$

*- $p < 0.05$, $(1-\beta)$ – Observed Power

Figure O.2 Two by two ANOVA for patellar tendon elongation (n=9).

2x2 Way ANOVA – $F_{(1, 8)} = 0.144$, $p = 0.714$, $(1-\beta = 0.06)$

↓

Main effects
Side - $F_{(1, 8)} = 1.170$, $p = 0.311$, $(1-\beta = 0.16)$
Session - $F_{(1, 8)} = 1.107$, $p = 0.324$, $(1-\beta = 0.15)$

$(1-\beta)$ – Observed Power

Figure O.3 Two by two ANOVA for patellar tendon strain (n=9).

2x2 Way ANOVA – $F_{(1, 8)} = 0.057$, $p = 0.817$, $(1-\beta = 0.06)$

↓

Main effects

Side - $F_{(1,8)} = 0.511$, $p = 0.495$, $(1-\beta = 0.10)$

Session - $F_{(1, 8)} = 0.432$, $p = 0.53$, $(1-\beta = 0.09)$

$(1-\beta)$ – Observed Power

Figure O.4 Two by two ANOVA for patellar tendon stress (n=9).

Table O.2 Two by two ANOVA ($F_{(\text{degrees of freedom})}$ test, p value and observed power $(1-\beta)$) for comparison of patellar tendon stiffness (10% intervals of maximal voluntary isometric contraction (MVIC)) between sessions (2) and sides (2) (n=9).

10%	F test		
interval	Session by Side	Main effect (session)	Main effect (side)
20 to 30	$F_{(1, 8)} = 0.654$, $p = 0.44_{1-\beta = 0.11}$	$F_{(1, 8)} = 1.05$, $p = 0.336_{1-\beta = 0.15}$	$F_{(1, 8)} = 1.72$, $p = 0.226_{1-\beta = 0.21}$
30 to 40	$F_{(1, 8)} = 1.86$, $p = 0.21_{1-\beta = 0.23}$	$F_{(1, 8)} = 1.47$, $p = 0.26_{1-\beta = 0.19}$	$F_{(1, 8)} = 1.90$, $p = 0.205_{1-\beta = 0.23}$
40 to 50	$F_{(1, 8)} = 5.487$, $p = 0.05^{*\dagger}{}_{1-\beta = 0.54}$		
50 to 60	$F_{(1, 8)} = 2.144$, $p = 0.18_{1-\beta = 0.25}$	$F_{(1, 8)} = 0.27$, $p = 0.619_{1-\beta = 0.07}$	$F_{(1, 8)} = 1.06$, $p = 0.333_{1-\beta = 0.15}$
60 to 70	$F_{(1, 8)} = 0.908$, $p = 0.37_{1-\beta = 0.13}$	$F_{(1, 8)} = 0.72$, $p = 0.422_{1-\beta = 0.12}$	$F_{(1, 8)} = 0.40$, $p = 0.55_{1-\beta = 0.09}$
70 to 80	$F_{(1, 8)} = 0.656$, $p = 0.44_{1-\beta = 0.11}$	$F_{(1, 8)} = 3.94$, $p = 0.083_{1-\beta = 0.42}$	$F_{(1, 8)} = 3.58$, $p = 0.10_{1-\beta = 0.38}$

*- $p < 0.05$, † - see Table O.3

Table O.3 One-way ANOVA ($F_{(\text{degrees of freedom})}$ test, p value and observed power $(1-\beta)$) for comparison of patellar tendon stiffness (10% interval from 40 to 50% maximal voluntary contraction) between sessions and between sides.

Comparison	$F_{(\text{degrees of freedom})}$ test, p value, $(1-\beta)$
Session 1 v Session 2 on Left	$F_{(1, 8)} = 2.933$, $p = 0.125$, $(1-\beta = 0.33)$
Session 1 v Session 2 on Right	$F_{(1, 8)} = 1.681$, $p = 0.433$, $(1-\beta = 0.11)$
Left v Right during Session 1	$F_{(1, 8)} = 5.727$, $p = 0.044^*$, $(1-\beta = 0.56)$
Left v Right during Session 2	$F_{(1, 8)} = 0.015$, $p = 0.709$, $(1-\beta = 0.06)$

*- $p < 0.05$

Table O.4 Two by two ANOVA ($F_{(\text{degrees of freedom})}$ test, p value and observed power $(1-\beta)$) for comparison of patellar tendon elastic modulus (10% intervals of maximal voluntary isometric contraction (MVIC)) between sessions (2) and sides (2) (N=13).

10%	F test		
interval	Session by Side	Main effect (session)	Main effect (side)
20 to 30	$F_{(1, 8)} = 0.413$, $p = 0.54_{1-\beta = 0.09}$	$F_{(1, 8)} = 0.001$, $p = 0.98_{1-\beta = 0.05}$	$F_{(1, 8)} = 1.302$, $p = 0.29_{1-\beta = 0.17}$
30 to 40	$F_{(1, 8)} = 0.049$, $p = 0.83_{1-\beta = 0.05}$	$F_{(1, 8)} = 0.008$, $p = 0.93_{1-\beta = 0.05}$	$F_{(1, 8)} = 2.291$, $p = 0.17_{1-\beta = 0.27}$
40 to 50	$F_{(1, 8)} = 634$, $p = 0.45_{1-\beta = 0.11}$	$F_{(1, 8)} = 0.189$, $p = 0.68_{1-\beta = 0.07}$	$F_{(1, 8)} = 12.472$, $p = 0.01^{*\dagger}{}_{1-\beta = 0.87}$
50 to 60	$F_{(1, 8)} = 0.076$, $p = 0.79_{1-\beta = 0.06}$	$F_{(1, 8)} = 1.448$, $p = 0.26_{1-\beta = 0.19}$	$F_{(1, 8)} = 1.132$, $p = 0.32_{1-\beta = 0.16}$
60 to 70	$F_{(1, 8)} = 0.007$, $p = 0.94_{1-\beta = 0.05}$	$F_{(1, 8)} = 2.121$, $p = 0.18_{1-\beta = 0.25}$	$F_{(1, 8)} = 0.184$, $p = 0.68_{1-\beta = 0.07}$
70 to 80	$F_{(1, 8)} = 3.584$, $p = 0.10_{1-\beta = 0.39}$	$F_{(1, 8)} = 6.917$, $p = 0.03^{*\ddagger}{}_{1-\beta = 0.64}$	$F_{(1, 8)} = 3.499$, $p = 0.10_{1-\beta = 0.38}$

*- $p<0.05$, † - see Table O.5, ‡ - see Table O.6

Table O.5 Pairwise comparisons (mean difference, SEM [95%CI], p value and observed power (1-β)) of patellar tendon elastic modulus (GPa) (40 to 50% maximal voluntary isometric contraction) between left and right sides.

Comparison	Mean Difference (SD)	Standard error of the mean [CI]	p	1-β
Sides	-0.048 (0.041)	0.014 [-0.079 to 0.017]	0.008*	0.87

*- $p < 0.05$

Table O.6 Pairwise comparisons (mean difference, SEM [95%CI], p value and observed power (1-β)) of patellar tendon elastic modulus (GPa) (70 to 80% maximal voluntary isometric contraction) between sessions 1 and 2.

Comparison	Mean Difference (SD)	Standard error of the mean [CI]	p	1-β
Sessions	0.292 (0.333)	0.111 [0.036 to 0.548]	0.03*	0.64

*- $p < 0.05$

2x2 Way ANOVA – $F_{(1, 8)} = 0.007$, $p = 0.936$, ($1-\beta = 0.05$)

↓

Main effects
Side - $F_{(1,8)} = 4.865$, $p = 0.058$, ($1-\beta = 0.49$)
Session - $F_{(1, 8)} = 8.837$, $p = 0.018$*, ($1-\beta = 0.74$) (see Table O.7)

*- $p < 0.05$, ($1-\beta$) – Observed Power

Figure O.5 Two by two ANOVA for percentage of patellar tendon work (50-100% MVIC) (n=9).

Table O.7 Pairwise comparisons (mean difference, SEM [95%CI], p value and observed power) of patellar tendon work (J) (50 to 100% maximal voluntary isometric contraction) between sessions 1 and 2.

Comparison	Mean Difference (SD)	Standard error of the mean [CI]	p	1-β
Sessions	-5.512 (5.562)	1.854 [-9.787 to -1.236]	0.018*	0.74
Sides	-4.835 (6.577)	2.192 [-9.891 to 0.220]	0.058	0.49

*- $p < 0.05$

Left 2-Way ANOVA – $F_{(1, 7)} = 0.287$, $p = 0.755$, $(1\text{-}\beta = 0.09)$

Main Effects
Session- $F_{(1, 7)} = 1.546$, $p = 0.254$, $(1\text{-}\beta = 0.19)$
Site - $F_{(2, 14)} = 4.071$, $p = 0.04$*, $(1\text{-}\beta = 0.62)$
(See Table O.8)

Left Multivariate test – $F_{(2. 6)} = 4.471$, $p = 0.065$, $(1\text{-}\beta = 0.531)$

Right 2-Way ANOVA – $F_{(2, 16)} = 0.146$, $p = 0.866$, $(1\text{-}\beta = 0.07)$

Main Effects
Session- $F_{(1, 8)} = 1.814$, $p = 0.215$, $(1\text{-}\beta = 0.22)$
Site - $F_{(2, 16)} = 5.129$, $p = 0.019$*, $(1\text{-}\beta = 0.74)$
(See Table O.8)

Right Multivariate test – $F_{(2, 7)} = 4.126$, $p = 0.066$, $(1\text{-}\beta = 0.53)$

*- $p < 0.05$, $(1\text{-}\beta)$ – Observed Power

Figure O.6 Two-way ANOVA comparing the patellar tendon cross-sectional area between sessions (one and two) and sites (proximal, mid and distal) for the left and right side (n=9).

Table O.8 Pairwise comparison (mean difference, SEM [95%CI] and p value) between proximal, mid and distal patellar tendon cross-sectional areas (mm^2) for the left and right sides.

Side	Comparison	Mean Difference	Standard error (95% CI)	p
Left	Proximal v Mid	5.09	2.21 (-1.829 to 12.004)	0.165
Left	Proximal v Distal	6.11	2.12 (-0.50 to 12.732)	0.071
Left	Mid v Distal	1.0	2.53 (-6.888 to 8.926)	1.00
Right	Proximal v Mid	8.42	2.741 (0.151 to 16.682)	0.046*
Right	Proximal v Distal	1.16	2.392 (-6.051 to 8.373)	1.00
Right	Mid v Distal	-7.26	3.332 (-17.305 to 2.794)	0.183

*- $p < 0.05$

Appendix P – Conference Presentation Abstracts

Comparison of patellar tendon mechanical properties in jumping and non-jumping elite athletes

Gupta A[1,2,3], Purdam C[3], Cook J[4], Allison GT[5]

[1]La Trobe University, Melbourne [2]Australian Institute of Sport, Canberra, [3]University of Sydney, Sydney, [4]Deakin University, Melbourne, [5]Curtin University & Royal Perth Hospital, Perth

The purpose of this investigation was to determine the inter-side and group differences in patellar tendon (PT) mechanical properties of elite athletes. Fifty one elite athletes (18 male volleyballers (MVB), 13 male basketballers (MBB), 14 female basketballers (FBB) and 6 swimmers (SW)) with normal healthy tendons were recruited. Calculation of PT force and mechanical properties (stiffness, strain, stress and modulus) were made following *in vivo* ultrasound measurement of PT elongation and, cross-sectional area with knee extension moment. Two-way repeated ANOVA revealed no inter-side difference in variables in jumping athletes while SW had less PT stress on the left than right (21.2 (4.7) V 25.3 (5.6) MPa) ($p < 0.02$) with a trend ($p = 0.05$) to be weaker on the left than right by 14. 6%. Between group comparison demonstrated MVB to experience greater PT stress than FBB (22.4 (7.6) v 15.5 (4.2) MPa) ($p = 0.03$). Groups were stratified by preferred landing and jumping leg and independent t-tests performed. This revealed that on the left side of MVB, those that prefer to land on their left leg had a lower PT stiffness than those who prefer to land on their right leg ($p < 0.03$). This may demonstrate a sport specific adaptive response in elite MVB based on landing leg preference when PT mechanical properties in jumping athletes are symmetrical. The PT may exhibit a strategy for maintaining a similar behaviour under load for all athletes regardless of the function to which the PT is exposed.

A comparison of two methods for calculation of vertical leg stiffness during hopping

Gupta A[1,2], Purdam C[3], Cook J[4], Allison GT[5]

[1]La Trobe University, Melbourne [2]Australian Institute of Sport, Canberra, [3]University of Sydney, Sydney, [4]Deakin University, Melbourne, [5]Curtin University & Royal Perth Hospital, Perth

The purpose of this investigation was to compare a gold standard method of calculating vertical leg stiffness (k_L) during hopping to a field test (FT) developed based on the model of a spring-mass model oscillating in simple harmonic motion (Dalleau et al, 2004). Nine subjects performed double-leg (DL) and alternate-leg (AL) hopping tasks on flat and decline surfaces at 2 frequencies (self-selected and 1.8 Hz). Peak vertical ground reaction force (F_{max}) and height change during the first half of contact (h_c) were derived from forceplate data. The FT uses flight (T_f) and contact (T_c) times to determine F_{max} ($F_{max} = (Mg.\pi/2)(T_f/T_c + 1)$) and h_c ($F_{max}/M(T_c^2/\pi^2) + gT_c^2/8$) where M is body mass and g is gravitational acceleration (9.81m/s^2). Comparison of methods for calculation of F_{max}, h_c and k_L for the 8 tasks revealed the main difference in values being due to the task being either bipedal or unipedal. Although there was a strong correlation (Pearson's r = 0.9) comparing the two methods for calculation of k_L, the FT method led to an overestimation of F_{max} for AL (1.7 times M) and underestimation in DL (0.6 times M) tasks. Similarly h_c was overestimated for all tasks and 2.6 times greater for AL than DL tasks. Stiffness was underestimated by 47.1% (DL) and 18.3% (AL) with a significant difference ($p < 0.001$) between DL and AL tasks for the FT method. The FT method provides an accurate value for k_L if the nature of the task is considered.

Evaluation of patellar tendon mechanical properties and the preferred landing leg in elite jumping athletes

Gupta A[1,2], Allison G.T.[3], Cook J.[4], and Purdam C[2]

[1]La Trobe University, Melbourne, Victoria, Australia, [2]Australian Institute of Sport, Canberra, Australia, [3]Curtin University, Perth, Western Australia, Australia, [4]Deakin University, Melbourne, Victoria, Australia

KEY WORDS: patellar tendon, mechanics, landing leg

INTRODUCTION: The purpose of this investigation was to evaluate patellar tendon (PT) mechanical properties in healthy, elite jumping athletes and the association with their preferred landing leg.

METHOD: Elite athletes (N=38) free of musculoskeletal pain were recruited from the Australian Institute of Sport, Australia. Subjects included 11 male volleyballers (MVB) (mean (SD) age (years), height (m) and weight (kg)) (17.5 (0.1), 2 (0.1) and 91.6 (8.9)), 13 male basketballers (MBB) (17.2 (1), 2 (0.1) and 87 (13)) and 14 female basketballers (FBB) (17 (0.8), 1.8 (0.1) and 78.1 (10.6)). Diagnostic ultrasound (US) (12MHz transducer, Nemio, Toshiba, Japan) was used to record the PT cross-sectional area. Ramped maximal voluntary isometric knee extension contraction (MVIC) at 90° knee flexion was performed (KinCom dynamometer, Chattanooga, USA). Knee torque and PT length, from inferior pole of the patella to the tibial tuberosity (Image J software http://rsb.info.nih.gov/ij/download.html), were determined at rest and 10% MVIC increments. PT force was calculated as knee extension torque divided by PT moment arm length, measured utilising a custom-made caliper (ICC = 0.7). PT mechanical properties (strain, stress, stiffness and elastic modulus) were calculated at 10% increments from 0 to 100% MVIC effort. T-tests were performed for inter-side and group comparisons. Sporting groups were stratified by landing leg as either left preferred landing leg (LPLL) or right preferred landing leg (RPLL).

RESULTS: In MVB the left and right PT are less stiff for the LPLL group compared to the left and right PT of the RPLL group ($p \leq 0.02$) only at sub-maximal efforts in MVB (Table 1).

Table 1 Male volleyball – Patellar tendon stiffness (MN/m) (Mean (SD))

	Left preferred landing leg (n=7)	Right preferred landing leg (n=4)
Left (40-60% MVIC) ($p < 0.02$)	0.83 (0.15)	1.16 (0.23)
Right (20-60% MVIC) ($p < 0.02$)	0.68 (0.10)	1.09 (0.37)

PT strain, stress, stiffness and elastic modulus were no different betweens sides for MVB, MBB or FBB, or between LPLL and RPLL groups in MBB (n=11 for LPLL and n=2 for RPLL) or FBB (n=8 for LPLL and n=6 for RPLL).

DISCUSSION: In volleyball, where there is a greater prevalence of jumper's knee than basketball (Lian Ø.B. et al, 2005), exposure to a sport-specific PT loading strategy may contribute to differences in PT mechanical properties based on the preferred landing leg.

CONCLUSION: This preliminary investigation may demonstrate sport-specific PT loading strategy observed as a difference in PT stiffness based on the preferred landing leg.

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Evaluation of patellar tendon mechanical properties in elite jumping athletes

Gupta A[1,2], Purdam C[1], Cook J[3], Allison G.T.[4]

[1]Australian Institute of Sport, Australia, [2]La Trobe University, [3]Deakin University, [4]Curtin University & Royal Perth Hospital

Introduction: The purpose of this investigation was to evaluate laterally in relation to patellar tendon (PT) mechanical properties in jumping athletes. ***Methodology*:** Forty-five athletes (14 male volleyballers (MVB), 17 male basketballers and 14 female basketballers) were recruited from the Australian Institute of Sport. *In vivo* PT properties (elongation and cross-sectional area) were measured utilising ultrasound (Nemio, Toshiba, Japan). Knee torque was measured using a dynamometer (Kincom, Chattanooga, USA) during a maximal voluntary isometric contraction (MVIC) of
the quadriceps. PT force was calculated by dividing torque by a calliper measure of PT moment arm length (ICC = 0.7). PT mechanical properties (strain, stiffness (K), stress and elastic modulus) were calculated at 10% increments of the MVIC to allow comparison between subjects. t-Tests were performed for inter-side and group comparisons. ***Results*:** Thirty-eight of 45 subjects had a normal PT bilaterally and were analysed. MVB with a left preferred landing leg (LPLL) had a left PT (LPT) K (MN/m) (0.83 (0.15)) lower than the LPT (1.16 (0.23)) of athletes with a right preferred landing leg (RPLL) ($p < 0.02$) at sub-maximal forces (40–60% MVIC). MVB with a LPLL had lower right PT (RPT) K (0.68 (0.1)) than the RPT (1.09 (0.37)) in those with a RPLL at sub-maximal forces (20–60% MVIC) ($p < 0.02$). Total and incremental PT mechanical properties were no different between sides for MVB, MBB or FBB. No difference between right or left preferred landing leg groups was found in either MBB or FBB. ***Conclusions*:** These findings may highlight a sport-specific PT loading strategy and will be discussed.

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