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Control of the lower limb during a fatiguing dynamic task

Kurt Laurence Mudie

16728439

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I love you.

Statement of Authenticity

The work presented in this thesis is, to the best of my knowledge and belief, original except as acknowledged in the text. I hereby declare that I have not submitted this material, either in full or in part, for a degree at this or any other institution.

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Kurt Laurence Mudie

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List of Publications and Abstracts

Peer-reviewed journal articles

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Peer-reviewed conference presentations

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Mudie, K. L., Gupta, A., & Clothier, P. J. (2015). *The effect of fatigue on lower limb motor variability during a controlled repetitive stretch-shorten cycle task.* Proceedings of the 33rd International Conference on Biomechanics in Sport, Poitiers, France, 29/6 - 3/7, 2015. (Appendix D)

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Abbreviations

3D	Three-dimensional
ACL	Anterior Cruciate Ligament
AMTI	Advanced Mechanical Technology, Inc.
ANOVA	Analysis of Variance
BF	Biceps Femoris
BNC	Bayonet Neill-Concelman
cm	Centimetre
COM	Centre of Mass
ΔCOM_f	Vertical Displacement of the Centre of Mass during the Flight Phase
ΔCOM_L	Vertical Displacement of the Centre of Mass during the Loading Phase
EMG	Electromyography
FL	Fibularis Longus
GM	Gluteus Medius
GRF	Ground Reaction Force
Hz	Hertz
iEMG	Integrated Electromyography
kg	Kilogram
K_{vert}	Vertical Stiffness
$\text{m}\cdot\text{s}^{-1}$	Meters per Second
M1	Short-latency Stretch Reflex
M2	Long-latency Stretch Reflex
M3	Second Long-latency Stretch Reflex
MAV	Mean Amplitude Value
MG	Medial Gastrocnemius
MHz	Megahertz
ms	Millisecond
mV	Millivolt
N	Newtons
NDI	Northern Digital Inc.

N·m·kg ⁻¹	Newton Meter per Kilogram
N·mm	Newton Millimetre
ODAU II	Optotrak Data Acquisition Unit II
OFM	Oxford Foot Model
ROM	Range of Motion
s	Second
SD	Standard Deviation
sEMG	Surface Electromyography
SENIAM	Surface Electromyography for the Non-Invasive Assessment of Muscles
SOL	Soleus
SSC	Stretch-Shorten Cycle
TA	Tibialis Anterior
vGRF	Vertical Ground Reaction Force
VL	Vastus Lateralis

Abstract

Human locomotion often requires the performance of submaximal, repeated and rhythmical efforts over an extended period of time. The ability to maintain consistent performance for extended periods is likely due to multiple adaptive mechanisms and may be affected by the progression of fatigue. However, it remains unclear whether changes in movement strategies are due to fatigue, changes in task performance or the repetitive nature of the task. Further, it is unknown if there are differences in movement strategies between the sexes during extended and fatiguing locomotive tasks. The aims of this thesis were to investigate the influence of fatigue and the effect of sex on neuromechanical characteristics during a repetitive and rapid lower limb loading task while performance output was maintained. To address these aims, a series of experimental investigations were completed.

On-the-spot, single-leg hopping was used to model the performance of a repetitive and rhythmical lower limb loading task. Hopping has been shown to have lower variability in performance output measures including cadence, displacement of the centre of mass (COM) and mechanical stiffness of the lower limb compared to other locomotor tasks such as walking or running. Therefore, hopping was deemed to be a task that would be more sensitive to changes in neuromechanical characteristics and better able to discern differences due to fatigue and sex.

Vertical stiffness (K_{vert}) has been used extensively to describe leg mechanical characteristics during terrestrial activity. However, there remains no gold standard method for the measurement of performance output during single-leg hopping. A preferred measure of K_{vert} was determined by assessing the level of agreement between two kinetic and two kinematic-kinetic methods used to calculate K_{vert} and displacement of the COM during single-leg hopping. Thirty eight participants completed a 10 s single-leg hopping effort on a force plate, with capture of 3D motion of the lower limb, pelvis and trunk. Key findings were that the segmental analysis method, which used the location and masses of the seven modelled segments to determine COM trajectory, had the least assumptions and demonstrated the highest repeatability for the calculation of vertical displacement of the COM and K_{vert} . This method had the greatest validity in determining K_{vert} and was therefore

deemed a preferred measurement method. Subsequently, the segmental analysis method was used to measure performance output.

The effect of local muscle fatigue on lower limb joint coupling variability and neuromuscular activity was investigated during exhaustive single-leg hopping to exhaustion when performance was maintained. Forty participants performed on-the-spot, single-leg hopping to volitional exhaustion at a set cadence of 132 hops per minute (audible metronome) and target hop height (tactile feedback). Performance output was maintained throughout the exhaustive trial with no significant change in K_{vert} , hopping frequency and the height of hopping. An increase in fatigue was confirmed by a significant decrease in post-trial maximal hop height performed after the exhaustive task, when compared to the pre-trial maximal hopping height. The main finding when comparing movement variability over the course of the fatiguing task was a significant increase in joint coupling variability of the knee-ankle and hip-knee flexion/extension axes and the knee flexion/extension-ankle eversion/inversion coupling as the task progressed. This finding demonstrated that the neuromotor system was able to adapt and maintain performance despite increasing fatigue. Further, this finding highlighted that during the performance of a rapid and repetitive loading activity, performance can be preserved with the utilisation of the available variability in the neuromotor system.

Movement is controlled by the synergistic action of muscles acting over the multi-segmented lower limb. Thus, adaptations in neuromuscular activity during the exhaustive single-leg hopping trial from the same participant pool were investigated. Muscle activation of the soleus (SOL) and vastus lateralis (VL) muscles significantly increased prior to landing and during the loading phase of the hop cycle, over the course of the exhaustive hopping trial. Changes in muscle activity were initially observed at the SOL muscle, and as fatigue progressed there was also an increase in muscle activity of VL. A significant increase in VL activity paralleled increases in joint coupling variability. The results from an investigation of movement variability and neuromuscular activity demonstrated a concordant temporal relationship between neural and mechanical characteristics as fatigue progressed. Greater contribution from synergist muscles such as the VL was observed during the single-leg hopping task which has been shown to primarily be modulated by triceps surae function. The greater contribution of knee extensors and increased movement

variability may represent an appropriate compensation strategy to maintain performance output and prolong the duration of hopping as local muscle fatigue increased.

To test the effect of sex on lower limb joint coupling variability and neuromuscular activity 20 men and 20 women completed a single-leg hopping task to exhaustion. Men and women maintained a similar performance output of K_{vert} , cadence and height of hopping over the duration of the task to exhaustion. However, women demonstrated a greater increase in VL muscle activity from the start to end of the task compared to men. Further, women demonstrated a greater increase in joint coupling variability from the start to the end of the task, with a convergence of values between men and women as the task progressed. These findings may further demonstrate that there was a relationship between changes in VL muscle activity and joint coupling variability. In addition, lower limb movement patterns were differentially modulated between sexes, with women demonstrating a greater increase in lower limb muscle activity and joint coupling variability than men as fatigue progressed during the continuous loading task.

The findings of these studies demonstrated a relationship between increased muscle activity and variability between adjoining lower limb segments as fatigue progressed. Further, men and women demonstrated differential changes as the level of fatigue increased. Women demonstrated a greater increase in joint coupling variability than men, which may have been associated with the greater relative dependence on the knee extensor muscle as the task progressed. These results extend the understanding of how there is an optimisation of variability in the neuromechanical system during the performance of repetitive tasks to fatigue in a healthy recreationally active population. Importantly, these results provide insight into what compensatory strategy may be utilised when motor performance was stereotypical.

CHAPTER ONE

Introduction

1.1 Background

Human locomotion requires the continuous and repeated loading of the lower limbs such as during walking, running and jumping. These tasks require a complex interaction of a multi-segmental skeletal system and synergistic action of muscles (Komi, 2000; Lamontagne & Kennedy, 2013; Nicol, Avela, & Komi, 2006). The maintenance of a consistent performance during locomotion and athletic activity is dependent on numerous intrinsic and extrinsic factors. Intrinsic control mechanisms that lead to performance of a motor task are dependent on many factors such as the central and peripheral regulation of neuromuscular activity (Farley, Houdijk, van Strien, & Louie, 1998; Farley & Morgenroth, 1999; Hobara, Kanosue, & Suzuki, 2007; Hortobágyi & DeVita, 2000), intersegmental coordination (Bartlett, Wheat, & Robins, 2007; Bernstein, 1967; Hamill, Palmer, & van Emmerik, 2012), optimal muscle strength (De Ruyter, van Leeuwen, Heijblom, Bobbert, & De Haan, 2006), muscle fibre type composition (Bosco & Komi, 1979) and tendon mechanical properties (Belli & Bosco, 1992). Therefore, execution of a motor task is susceptible to deterioration of one or more of these components.

Locomotion is most often performed at a submaximal effort in a repetitive and cyclical manner. Factors that have been attributed to affecting performance have been fatigue (Enoka & Duchateau, 2008; Gandevia, 2001; Nicol et al., 2006), sex (Thomas & French, 1985), current or previous injury (Bahr & Krosshaug, 2005; Hamill et al., 2012) and skill level (Bernstein, 1967; Newell, 1985). There is evidence to demonstrate that fatigue affects both the central and peripheral nervous system's ability to activate the motoneuron pool at the muscle (Gandevia, 2001). As an individual performs a progressively fatiguing task, there may be changes to performance such as how the movement is performed over the total duration of the task (Bonnard, Sirin, Oddsson, & Thorstensson, 1994; Morio, Nicol, Barla, Barthèlemy, & Berton, 2012; Moritani, Oddsson, & Thorstensson, 1990; Padua et al., 2006). It is important to understand how these factors that affect performance interact with the multiple intrinsic control mechanisms that produce movement.

Fatigue during repetitive lower limb loading tasks has been shown to affect multiple movement and muscle activation characteristics (Bonnard et al., 1994; Morio et al., 2012; Moritani et al., 1990; Padua et al., 2006). It is evident that a result of fatigue is

the cessation of task performance, which is defined as exhaustion. However, it remains unclear how movement changes during the progression of a task to exhaustion and what characterises a change in movement pattern.

Variability in human movement has been described to represent the available degrees of freedom, such that a decrease or redundancy in the degrees of freedom may result in a reduction in movement variability (Bernstein, 1967; Davids, Bennett, & Newell, 2006; Latash, 2000; Newell & Corcos, 1993). In the musculoskeletal system there are an inherently large number of muscles and segments that need to interact. Therefore, a multitude of opportunities for variability in the performance of human movement may be observed. It is suggested this abundance of potential movement strategies helps the system deal with unexpected perturbations and the ability to adapt to a variety of tasks. Thus, it is important to assess potential changes in coordination between multiple segments and the associated variability.

It has been suggested there is an “optimal” level of variability required for maintenance of healthy function (Davids et al., 2006; Hamill et al., 2012; Newell & Corcos, 1993; Preatoni et al., 2013; Stergiou, Harbourne, & Cavanaugh, 2006). Large increases or decreases in variability may be detrimental to the musculoskeletal system and associated with pathology or reduced performance (Bartlett et al., 2007; Hamill et al., 2012; Moraiti et al., 2009; Moraiti, Stergiou, Vasiliadis, Motsis, & Georgoulis, 2010; Preatoni et al., 2013). For example, large decreases below this level of optimal movement variability may lead to a rigid system whilst increases above this optimal level could mean a random and unstable system. The optimal level of variability may protect the musculoskeletal system from an increased risk of further injury and allow the maintenance of performance during a continuous task. However, it is unclear how fatigue influences movement variability and whether any observed compensation occurs during a continuous repeated loading task.

Studies have reported increased variability during treadmill walking (Ferber & Pohl, 2011), running (Meardon, Hamill, & Derrick, 2011), side-step manoeuvre (Cortes, Onate, & Morrison, 2014) and vertical jump task performed to fatigue (Dal Pupo, Dias, Gheller, Detanico, & Santos, 2013). In contrast, there have been studies to fatigue that have demonstrated a decrease in variability during a side-cut manoeuvre (Samaan, Hoch, Ringleb, Bawab, & Weinhandl, 2015) and sprinting task to fatigue

(Trezise, Bartlett, & Bussey, 2011). The conflicting findings may be due to variability being sensitive to either differences in tasks between studies or changes in task performance within the study that are specific to the fatigue protocol and movement task (Nicol et al., 2006; Santamaria & Webster, 2010). Therefore, it is necessary to investigate the link between movement variability and fatigue during a task where performance is controlled.

Multi-segment coordination in the lower limb requires an interaction between both the central nervous system and peripheral nervous systems to produce an appropriate motor outcome (Farley et al., 1998; Farley & Morgenroth, 1999; Hobara et al., 2007; Hortobágyi & DeVita, 2000). Observation of a change in movement patterns are likely due to adaptations in both central and peripheral components of the nervous system with a change being observed in the latency and amplitude characteristics of the electric signal at the muscles (Samaan et al., 2015; Srinivasan & Mathiassen, 2012). However, there are conflicting findings in muscle activity with reports of both an increase (Bonnard et al., 1994; Oliver, De Ste Croix, Lloyd, & Williams, 2014; Padua et al., 2006; Regueme, Nicol, Barthélemy, & Grélot, 2005) and decrease (Kuitunen, Avela, Kyröläinen, Nicol, & Komi, 2002; Kuitunen, Kyröläinen, Avela, & Komi, 2007; Morio et al., 2011; Moritani et al., 1990) in lower limb muscle activity following repetitive loading tasks to fatigue. In addition, the relationship between changes in neuromuscular activity and movement variability during a fatiguing task is yet to be investigated.

Although differences in performance output have been well documented between men and women, it is unclear if there are differences in movement strategies between the sexes during a fatiguing repetitive loading task. Women have been shown to have lower levels of movement variability than men during non-fatiguing dynamic tasks such as running or a side-cut manoeuvre (Barrett, Noordegraaf, & Morrison, 2008; Breen, 2012; Maulder, 2011; Pollard, Heiderscheit, van Emmerik, & Hamill, 2005). Further, recent reviews have reported mixed findings of between sex differences in neuromuscular temporal and amplitude activity during running, jumping, landing or side-cut tasks in non-fatiguing conditions (Benjaminse, Gokeler, Fleisig, Sell, & Otten, 2011; Bruton, O'Dwyer, & Adams, 2013). During a fatiguing side-cut task there were no between sex differences in muscle temporal and amplitude characteristics (Gehring, Melnyk, & Gollhofer, 2009). In contrast, compared to men,

women had greater quadriceps and tibialis anterior muscle activity during a fatiguing hopping task (Padua et al., 2006) and a delayed onset of the hamstring and vastus lateralis activity during a fatiguing landing task (Iguchi, Tateuchi, Taniguchi, & Ichihashi, 2014). Therefore, whether sex affects observed changes in variability and neuromechanical characteristics across the duration of a continuous task is unknown.

The contrasting changes in neuromechanical characteristics during a fatiguing task are likely due to task and time dependent factors. Changes in performance output following a fatiguing protocol such as decreased hop or jump height will likely be associated with changes in underlying neuromechanical measures. Further, task dependent factors shown to alter fatigue effects have demonstrated that tasks only containing a landing component without a subsequent push off phase have displayed differences in kinetics and kinematics (Edwards, Steele, & McGhee, 2010). Whereas for time dependent factors, there is evidence that changes in neuromuscular activity are non-linear over the duration of a fatiguing task (Meardon et al., 2011; Morio et al., 2011; Regueme et al., 2005). For example, the amplitude of soleus muscle activity was shown to decrease and increase in the first and last 25% of the duration an exhaustive sledge rebound task, respectively (Regueme et al., 2005). As the majority of studies have only investigated changes pre and post a fatiguing task, the progressive changes due to fatigue during a repetitive loading task remain relatively unknown. Therefore, a suitable task is needed to minimise confounding task and time dependent factors during a progressive fatiguing trial to ensure appropriate experimental design to investigate the effects of fatigue on movement variability and neuromechanical characteristics.

Single-leg hopping is a task that is relatively simple, is able to be controlled in a laboratory setting, and therefore limits confounding variables that may affect performance, such as differences in skill level and environmental conditions. Hopping allows investigation of kinetic, kinematic and muscle activation characteristics during a repetitive and rapid loading task to fatigue to better understand the interaction between these intrinsic control mechanisms that produce movement. Key performance output characteristics of hopping such as cadence, vertical height displacement and leg mechanical stiffness were able to be maintained during efforts of single-leg hopping to volitional exhaustion (Gupta, Hilliard, Mudie, & Clothier, 2016). This contrasts other tasks such as walking, running and repeated

jump-land tasks that also involve multi-segment coordination, and are susceptible to a greater magnitude of change in task performance compared to on-the-spot hopping (Lamontagne & Kennedy, 2013). Thus, by being able to control for performance characteristics, on-the-spot hopping allows for stronger conclusions to be made about changes in the biomechanical and physiological properties due to fatigue rather than other confounding task or time dependent factors.

During single-leg hopping, performance characteristics of cadence and ground reaction force are measured directly from a force platform. However, it is problematic to accurately measure the vertical displacement of the COM and vertical stiffness (K_{vert}). Consequently, a number of methods have been developed to estimate the displacement of the COM to calculate K_{vert} during double-leg hopping (Butler, Crowell III, & Davis, 2003; Hall, 2007; Hébert-Losier & Eriksson, 2014; Hobara, Inoue, Kobayashi, & Ogata, 2014; Ranavolo et al., 2008; Yang & Pai, 2014). However, there is a significant difference in K_{vert} between single- and double-leg hopping tasks (Brauner, Sterzing, Wulf, & Horstmann, 2014; Hobara, Kobayashi, Kato, & Ogata, 2013). The differences in magnitude and variability between computational methods for the calculation of K_{vert} may therefore be more or less pronounced in single-leg compared to double-leg hopping. Thus, a preferred method of measuring K_{vert} during single-leg hopping remains unknown.

1.2 Thesis outline

The aim of this thesis was to determine the effect of local muscle fatigue on the magnitude and variability of neuromechanical characteristics and whether these changes were similar between sexes when maintaining performance of a continuous dynamic lower limb loading task. Changes within subjects and between the sexes were evaluated to achieve the research aims.

The aims of the thesis were achieved by conducting a series of studies. A key feature in determining whether there were changes to neuromechanical characteristics during hopping to exhaustion was to maintain performance output characteristics, specifically K_{vert} . The first study was performed to determine a preferred method of measuring K_{vert} during single-leg hopping to ensure performance was maintained. The subsequent experimental work then evaluated the effect of exhaustion on neuromechanical characteristics and the differences in neuromechanical

characteristics between the sexes when performance was maintained. The findings from addressing each of the aims have been presented in four chapters for clarity. To minimise overlap in methodologies between experiments, the reader is referred to previous chapters where appropriate.

In summary, Chapter 2 reviewed the literature pertaining to changes in movement variability, neuromuscular activity and sex differences during fatiguing dynamic lower limb loading tasks. Contrasting findings and the gaps in the literature were highlighted, forming the foundation for the research studies in this thesis. First, prior to commencement of the experimental studies, a methodological study to determine a repeatable and preferred measure of performance output was performed. The agreement between K_{vert} calculated using two kinetic and two kinematic-kinetic measures of the vertical displacement of the centre of mass (COM) during the single-leg hopping task was assessed (Chapter 3). Chapter 4 examined the effect of local muscle fatigue on movement variability when performance output was maintained during a dynamic lower limb loading task. Chapter 5 built on the findings of Chapter 4 and examined the effect of local muscle fatigue on neuromuscular activity when performance output was maintained. Chapter 6 tested the effect of sex on lower limb movement variability and neuromuscular activity during a fatiguing dynamic lower limb loading task. Chapter 7 summarised the key findings of Chapters 3 – 6. The results of each chapter were discussed within a single coherent framework, addressing the overarching aim of the thesis. Limitations and future research directions were then discussed.

CHAPTER TWO

Changes in lower limb movement variability and neuromechanical characteristics due to fatigue and sex during dynamic tasks: A narrative review of literature

2.1 Introduction

The ability to maintain a consistent performance output during a continuous and repetitive task is important in many activities. During human locomotion, as fatigue progresses, maintaining a consistent performance is dependent on the interaction between the active and passive components of the muscle and the nervous systems that combine to form the musculoskeletal system. Variability is inherent in a biological system and considered a normal behaviour in movement. However, it is not clear how movement variability is affected by the performance of exhaustive tasks. Specifically, it is not clear whether observations of any change in movement patterns are in fact due to increasing levels of fatigue or due to sensorimotor integration during the performance of a repetitive task. The purpose of this review was to explore the factors that may affect movement variability and changes in neuromechanical characteristics due to fatigue and sex during dynamic tasks that were performed to exhaustion.

A narrative review was completed using the Google Scholar, Scopus, PubMed and Western Sydney University Library databases to identify key studies and reviews related to movement variability and neuromechanical characteristics during dynamic tasks, and the effect of fatigue and sex on these characteristics. Reference lists of all included articles were searched manually to identify further studies.

2.2 Movement variability

Movement variability is an inherent characteristic during task performance and is due to the abundant degrees of freedom of the multi-segmental system (Bernstein, 1967). As a greater understanding has developed to how lower limb movement is controlled, so too has the understanding of how variability of movement may be detrimental or protective of the musculoskeletal system. Although completing repeated performances may lead to an increase in skill of the performer, the observation of consistency or invariant movement is an interpretation of a decrease in movement variability (Bernstein, 1967; Latash, 2000, 2012; Newell & Corcos, 1993). Historically, variance within movement has been considered undesirable and simply artefactual as a measure during the performance of the task. In contrast recent motor control theories such as the dynamic systems theory suggests that variability serves a

functional purpose for the successful execution of a task (Bernstein, 1967; Hamill et al., 2012; Latash, 2012). Specifically, the dynamic system theory suggests that an optimal level of variability may protect the musculoskeletal system from an increased risk of injury or allow the maintenance of a consistent task performance.

Movement variability has been studied as a global measure of variability (end-point variability) and as part of the linked kinetic chain of the multi-segmented system (movement variability) (Bartlett et al., 2007; Davids et al., 2006; Hamill et al., 2012; Newell & Corcos, 1993; Preatoni et al., 2013). End-point variability refers to variance within the goal or performance characteristics, such as sprint time, target accuracy or stride rate (Hamill et al., 2012; Preatoni et al., 2013). Expert performers and healthy individuals typically have lower levels of variance in their performance output (end-point variability) than their less experienced or injured counterparts. For example, consistent horizontal block speed and 10 m sprint times within and between sessions is an essential feature of an elite sprinter (Bradshaw, Maulder, & Keogh, 2007). Further, individuals with Parkinson's Disease or those identified to be at an increased risk of falls, had higher levels of variability in their stride rate and stride length during walking compared to healthy individuals (Blin, Ferrandez, & Serratrice, 1990; Hausdorff, Rios, & Edelberg, 2001; Maki, 1997). These results demonstrate that there may be lower levels of end-point variability in both trained as well as healthy populations when compared to untrained or pathological populations, respectively (Hamill et al., 2012; Latash, 2012; Latash, Levin, Scholz, & Schöner, 2010).

Movement variability may be characterised by the changes in kinetics, kinematics and muscle activation patterns (Latash, Scholz, & Schöner, 2002). It has been demonstrated that variability of the intrinsic characteristics does not explicitly result in variability of the accompanying performance output (Bartlett et al., 2007; Newell & Corcos, 1993; Preatoni, Ferrario, Donà, Hamill, & Rodano, 2010; Preatoni et al., 2013). For example, expert blacksmiths demonstrated high variability in movement variability (joint angles) with low variability in end-point variability (hammer head trajectory) (Latash, 2012). Greater movement variability has been suggested to reflect more complex movement patterns observed in healthy people as well as a sign of well-developed motor skills (Stergiou & Decker, 2011). Such changes in movement variability may reflect the adaptability of the musculoskeletal system to

changes in response to imposed internal and/or external constraints (Stergiou & Decker, 2011).

Changes in variability appear to be context-specific. Low variability is important in performance output features of a successful task, i.e. sprint time or stride characteristics. Conversely, movement variability may be an important feature of a complex and healthy system, permitting adaptability to potential changes in task constraints. A decrease or increase in variability may be observed as compensations and both allow the continuation of a performance.

Research investigating the relationship between movement variability and injury risk supports the concept that there needs to be an optimal range of variability (Davids et al., 2006; Hamill et al., 2012; Hamill, van Emmerik, Heiderscheit, & Li, 1999; Maulder, 2011; Newell & Corcos, 1993; Preatoni et al., 2013; Stergiou & Decker, 2011; Stergiou et al., 2006). An association has been shown between lower limb injuries and relatively low levels of movement variability during continuous running (Hamill et al., 1999; Heiderscheit, Hamill, & van Emmerik, 2002; Herb et al., 2014; Miller, Meardon, Derrick, & Gillette, 2008; Seay, van Emmerik, & Hamill, 2011), jumping (Brown, Bowser, & Simpson, 2012) and cutting tasks (Edwards, Brooke, & Cook, 2016). Specifically, women with patellofemoral pain syndrome reported a decrease in variability at the thigh-shank coupling during treadmill running compared to control subjects (Heiderscheit et al., 2002). People with chronic ankle instability reported a decrease in variability of ankle frontal plane motion (Terada et al., 2015) and shank-rearfoot coupling during walking compared to control subjects (Herb et al., 2014). Further, knee and hip joint variability during single-leg jump landings in men and women with chronic ankle instability were lower than healthy controls (Brown et al., 2012). The decrease in variability observed in these studies may be a demonstration of a compensatory strategy to stiffen the joints by reducing the variability of the segment couplings. These cross-sectional studies may highlight a protective strategy that may have led to an avoidance of exposure to positions or joint loads, which may increase symptoms or expose the limb to further injury. The relative guarding has been proposed as a mechanism in people with symptoms and in people who have had a resolution of symptoms (Seay et al., 2011), which demonstrates the possible habituation of altered movement patterns following injury.

Cross-sectional studies do not establish relatively low levels of movement variability as causative of an increased risk of injury. However, lower variability may minimise the available degrees of freedom and limit the ability of the system to adapt to internal or external perturbations (Bartlett et al., 2007; Hamill et al., 2012; Preatoni et al., 2013), potentially exposing certain musculoskeletal structures to a greater cumulative load (Bartlett et al., 2007; Hamill et al., 1999; Meardon et al., 2011; Preatoni et al., 2013). Therefore, it is possible that observations of lower variability of segment couplings may contribute to maladaptive tissue damage or be a result of injury and act as a protective strategy.

Greater variability in stride time, lower limb kinematics and joint coupling variability has been observed in sample populations with pathology during treadmill walking (Moraiti et al., 2009; Moraiti et al., 2010), treadmill running (Hein et al., 2012) and overground running (Meardon et al., 2011). Further, relatively high and low levels of ankle joint variability were observed in participants with a lower limb pathology during double leg landings (James, Dufek, & Bates, 2000). In agreement, high and low levels of ankle, knee, hip and trunk joint variability were observed in Australian football players with a previous history of groin pain during a side-cut task (Edwards et al., 2016), as well as in lower limb joint coupling variability of previously injured participants during a land and cut task (Breen, 2012) and treadmill running (Miller et al., 2008). The contrasting findings of either decreased or increased variability in participants with pathology may suggest that there is a fluctuating range of variability that is dependent on the task performed, the dependent variable calculated and the sample populations evaluated. Further, differences in variability may be due to adaptations along the kinetic chain such that low variability at one segment may be compensated by high variability elsewhere (Winter, 1984).

Based on the retrospective study design used in the previous investigations (Edwards et al., 2016; Hamill et al., 1999; Heiderscheit et al., 2002; Herb et al., 2014; James et al., 2000; Meardon et al., 2011; Miller et al., 2008; Seay et al., 2011), it is not possible to determine the cause-effect relationship between movement variability and injury. Therefore, it is unclear whether changes in movement variability caused the increased risk of injury or whether the increased risk of injury caused the changes in movement variability. To date, only a single cohort study by Maulder (2011) has been undertaken to demonstrate a possible cause and effect relationship between

lower limb variability and injury in which relatively high and low levels of joint coupling variability were found to be associated with injury occurrence. Specifically, the authors reported low joint coupling variability was associated with injury occurrence in women, and high joint coupling variability was associated with injury occurrence in men. The cohort design of this study with the findings of a possible association between changes in movement variability in numerous cross-sectional studies does highlight the theory of “optimal” movement variability and suggests a link between changes in variability and increased risk of injury.

As a phenomenon, optimal movement variability has been shown to be associated with protecting and contributing to injury (Bartlett et al., 2007; Hamill et al., 1999; Meardon et al., 2011; Preatoni et al., 2013). This has been reflected in previous research demonstrating both healthy and skilled individuals to reflect some degree of variance within and between their performances. Large decreases in movement variability may lead to a rigid system whilst increases could mean an unstable system. High or low variability may lead to an inability to effectively adapt to changes in the external environment possibly leading to an increased risk of acute or overuse injury during locomotion and sporting tasks (Bartlett et al., 2007; Hamill et al., 2012; Hamill et al., 1999; Moraiti et al., 2009; Moraiti et al., 2010; Preatoni et al., 2013). A dynamic systems approach suggests the investigation of variability in human movement can provide insight into the function, control and adaptability of the system (Seifert, Button, & Davids, 2013; van Emmerik, Ducharme, Amado, & Hamill, 2016). Based on the current understanding of the association between the health of the system and changes in movement variability, future investigations into how variability during dynamic movements adapts under various conditions and constraints is required.

2.2.1 Variability and skill level

Observations of the intrinsic features of human movement have suggested that variability within these movement patterns is crucial for maintenance of a consistent performance (Bernstein, 1967; Hamill et al., 2012; Latash, 2012). This variability within a movement pattern may be particularly evident in tasks that have multiple degrees of freedom such as running and jumping, which require a coordinated movement strategy between the segments (Davids et al., 2006; Stergiou, 2004).

Further, a number of studies have demonstrated that variability in neuromechanical characteristics are different between participants with different levels of skill (Chapman, Vicenzino, Blanch, & Hodges, 2008, 2009; Sides & Wilson, 2012; Wilson, Simpson, van Emmerik, & Hamill, 2008). The higher skilled international race walkers displayed greater variability in their vertical ground reaction force (vGRF) and sagittal hip and ankle angle excursions during the stance phase when compared to the lesser skilled national level race walkers (Preatoni et al., 2010). During the overground race walking trial the regularity and predictability of changes in the neuromechanical characteristics were measured by sample entropy (Preatoni et al., 2010). However, the relationship between the level of skill and coordination variability such as joint coupling variability has been shown to be non-linear. This was demonstrated when comparing five elite triple jumpers (Wilson et al., 2008). The highest skilled performers had greater joint coupling variability in sagittal knee and hip angles of the stance leg during the transition of the hop-step phase compared to the intermediate performers (Wilson et al., 2008). However, the lowest skilled performers also had greater joint coupling variability than the intermediate performers.

The relationship between variability and skill level is also dependent on the movements and complexity of the task. The findings in triple-jumpers contrasts those of elite level cyclists who were shown to have lower movement variability of the sagittal hip-ankle and knee-ankle motion (Chapman et al., 2009; Sides & Wilson, 2012), muscle amplitudes and co-activation ratios of lower limb muscles compared to a group of matched novices (Chapman et al., 2008). As opposed to running and jumping, cycling mechanically constrains the athlete into a relatively fixed position limiting the joint excursions, forces and velocities in which the lower limb may be functional. This suggests that skilled performers in elite level cycling, due to the nature of the task, have greater redundancy in the available degrees of freedom compared to novices. The contrasting results with trained athletes demonstrating both higher (Preatoni et al., 2010; Wilson et al., 2008) and lower levels of variability (Chapman et al., 2008, 2009; Sides & Wilson, 2012) compared to relatively untrained individuals may be explained due to the tasks which have been evaluated, highlighting the ability of the neuromotor system to adapt differently based on the demands of the task. Coordination of the control parameters that contribute to

movement emerge from imposed constraints by internal and external factors, including the task, environmental dynamics and the individual (Newell, 1985). Therefore, any observed movement variability may be accounted for by a combination of the performer's skill level and due to the difficulty and complexity of the task being performed. The focus of the current research project was to evaluate the influence of fatigue and sex on neuromechanical characteristics and eliminate the influence of confounding factors such as changes in task performance or environmental constraints.

The level of skill directly influences performance as has been demonstrated in sports such as cycling (Chapman et al., 2008, 2009; Sides & Wilson, 2012), running (Preatoni et al., 2013), walking (Preatoni et al., 2010) and jumping (Preatoni et al., 2013; Wilson et al., 2008). For a specific task, this may be illustrated as a learning effect (Figure 2.1) in which the greater the skill level, then the greater the performance. However, in the early stages of learning there is a relatively greater learning effect compared to learning towards mastery (Fitts & Posner, 1967). High variability during the early stages of learning may be experienced by the novice who explores different movement patterns, has more random movements and thus a large range of movement variability (Fitts & Posner, 1967). As skill acquisition progresses there may be a decrease in the degrees of freedom (decreased variability) resulting in greater accuracy and probable reduction in the range of intrinsic movement patterns, demonstrated by the decrease in variability from untrained to trained individuals (Chapman et al., 2008, 2009; Sides & Wilson, 2012). As learners reach expertise, there may be an increase in movement variability providing flexibility in how a task is performed with minimal or no change in performance measures such as accuracy or speed. This notion is supported by the increase in variability from sub-elite to elite level athletes (Preatoni et al., 2010; Wilson et al., 2008). Therefore, movement variability is inherently dependent on the level of the skill of the performer and may be represented by an inverse-bell shaped curve (Figure 2.1) (Seifert et al., 2013; Stergiou & Decker, 2011). Supporting an inverse bell shaped curve, a theoretical model by Stergiou & Decker (2011) illustrated more complex movement patterns to exhibit a blend of predictability and variability (randomness). These movements can be described as "repetition without repetition" (Bernstein, 1967; Latash, 2012), suggesting there may be a certain magnitude of variability between consecutive

repetitions yet the same successful outcome still achieved (Latash et al., 2002; Seifert et al., 2013).

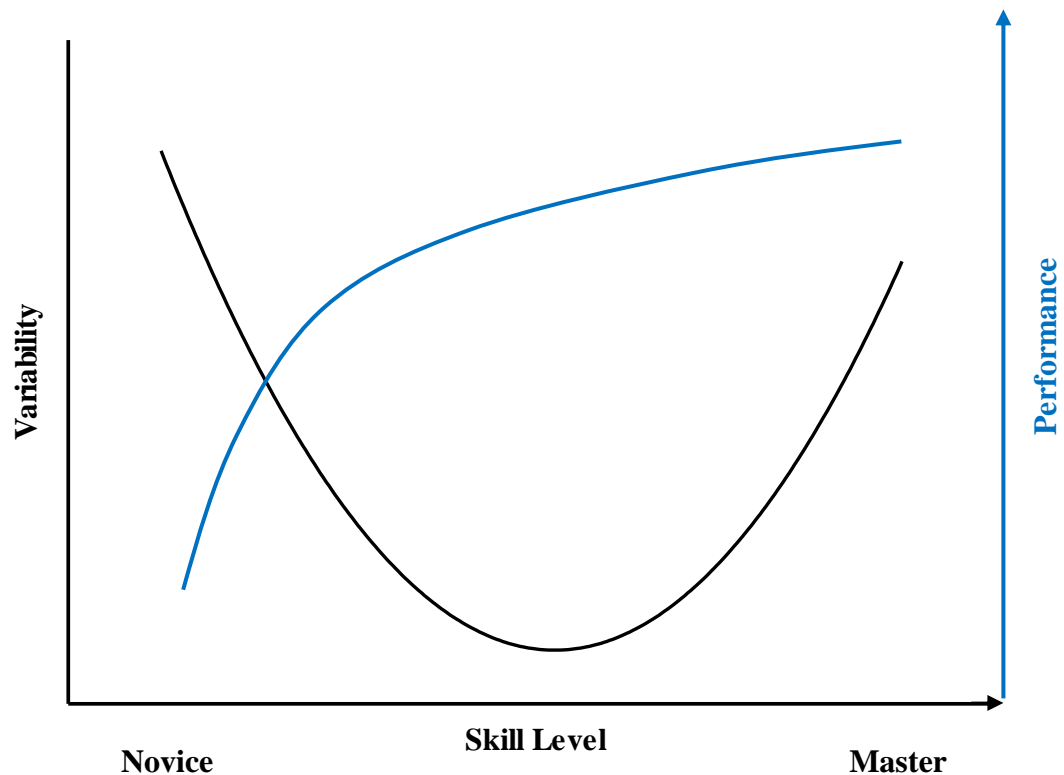


Figure 2.1 A schematic illustrating a motor learning effect from novice to mastery level of skill acquisition, representing a non-linear hypothesis for the relationship between variability (black line), performance (blue line) and skill level, adapted from (Stergiou & Decker, 2011). The novice (far left on the x -axis) initially decreases movement variability when approaching the intermediate skill stage (middle of the x -axis) eliminating the apparent redundant degrees of freedom. As the intermediate performer approaches an elite level (far right of the x -axis) they increase their available degrees of freedom to improve adaptability and flexibility to potential perturbations.

Overall, the dynamical systems theory considers within-subject variability as being necessary to optimise performance and adaptability within the ever changing environment (Bartlett et al., 2007; Hamill et al., 1999; Preatoni et al., 2013). This is necessary during continuous and repetitive tasks during which a narrow range of variability may be appropriate if there are no intrinsic or extrinsic perturbations. However, considering that environmental factors are fluctuant and that there are

intrinsic changes as a task progresses such as fatigue, there may be a need for greater movement variability to achieve an appropriate performance outcome.

2.3 Fatigue and movement variability

Muscle fatigue is defined as the reduction in muscular force or power, regardless of the force required and whether or not the task can be maintained (Bigland-Ritchie & Woods, 1984; Enoka & Duchateau, 2008). This describes fatigue as a continuum, during which there are greater changes to the levels of cognitive and physiological fatigue as a continuous tasks progresses (Gandevia, 2001; Nicol et al., 2006). The inability to maintain the required performance output has been termed volitional exhaustion (Morio et al., 2011; Nicol et al., 2006; Regueme et al., 2005). This is typically the end-point of progressive fatigue, whereby there is failure to continue with the motor task (Morio et al., 2011; Nicol et al., 2006; Regueme et al., 2005).

Daily activities that are often exposed to a varying level of fatigue such as walking, running and jumping, commonly utilise a stretch-shorten cycle (SSC) muscle action. The SSC is defined as a muscle which is first actively stretched (eccentric phase) and then rapidly shortened (concentric phase; Komi, 2000; Nicol et al., 2006). The successful execution of these SSC tasks requires a complex interaction of muscle actions to produce coordinated multi-segmental motion (Bernstein, 1967; Lamontagne & Kennedy, 2013; Lloyd, Oliver, Hughes, & Williams, 2012). Fatigue during repetitive tasks has been shown to affect multiple system outputs including the kinetics, kinematics and neuromuscular characteristics during a task (Bonnard et al., 1994; Morio et al., 2012; Moritani et al., 1990; Padua et al., 2006). Due to the rapid, repetitive and often high load requirements of sporting tasks that repetitively utilise the SSC compared to activities of daily living, there may be a failure to optimise the absorption of forces during loading as fatigue progresses. It has been suggested that suboptimal coordination of neuromechanical characteristics and movement variability during dynamic tasks may contribute to an increased risk of sustaining acute and overuse lower limb musculoskeletal injuries (Bartlett et al., 2007; Hamill et al., 2012; Hamill et al., 1999; Moraiti et al., 2009; Moraiti et al., 2010; Preatoni et al., 2013; Santello, 2005).

There are conflicting findings of changes in movement variability during repeated loading tasks to fatigue (Cortes et al., 2014; Dal Pupo et al., 2013; Edwards, Steele,

Cook, Purdam, & McGhee, 2012; Ferber & Pohl, 2011; Meardon et al., 2011; Miller et al., 2008; Samaan et al., 2015; Trezise et al., 2011). During a continuous, overground running task to exhaustion, Meardon et al. (2011) demonstrated an increase in variability of stride time as fatigue progressed when evaluated using detrended fluctuation analysis (Meardon et al., 2011). Interestingly, the authors demonstrated the greatest increase in stride time variability during the first half compared to the second half of the run when there was no significant change in stride time variability (Meardon et al., 2011), suggesting the greater change in the first part of the run was a motor learning affect rather than due to fatigue. Variability of knee frontal plane angles during a sidestep cutting task after a fatiguing protocol were found to increase when evaluated using a change in sample entropy, which measured the repeatability of time series data (Cortes et al., 2014). Further, the coefficient of variation of ankle and knee joint range of motion during sledge rebounds increased in the fatigued compared to non-fatigued condition (Edwards et al., 2012). Similarly, joint coupling variability increased for the trunk-thigh couplings in the sagittal plane during a fatiguing bout of repetitive maximal vertical jumps for 30 s (Dal Pupo et al., 2013). Furthermore, there was an increase in shank-rearfoot and rearfoot-forefoot couplings during treadmill walking following isolated fatigue of the tibialis posterior muscle (Ferber & Pohl, 2011). Dal Pupo et al., (2013) and Ferber & Pohl, (2011) quantified the coordination between two oscillating segments using two different methods, a continuous relative phase analysis and vector coding, respectively. Whilst continuous relative phase and vector coding techniques provide similar information, caution should be taken when directly comparing findings from these two different analysis methods (Miller, Chang, Baird, van Emmerik, & Hamill, 2010). Overall, these studies demonstrate a relationship between fatigue and increased levels of movement variability.

Whilst increases in joint coupling variability with fatigue has been shown (Dal Pupo et al., 2013; Ferber & Pohl, 2011), hip-knee joint coupling variability decreased during the impact and weight acceptance phases of a sidestep cutting task following an isolated hamstring fatigue protocol (Samaan et al., 2015). Variability of the hip-knee coupling also decreased in two sprinters after completing six 100 m maximal effort sprints although when compared between the start and end of the 100 m trials one sprinter decreased while the other increased hip-knee coupling variability

(Trezise et al., 2011). These findings also demonstrate a relationship between increased levels of fatigue and a reduction in movement variability.

Previous studies conflict in the type of changes of movement variability during continuous tasks which may be influenced by a number of factors. For example, protocols to induce fatigue vary and have included efforts that achieve complete exhaustion (Ferber & Pohl, 2011; Meardon et al., 2011; Miller et al., 2008; Samaan et al., 2015), while other studies have used protocols with a fixed amount of work or effort (Dal Pupo et al., 2013; Trezise et al., 2011). Therefore, it is plausible that different protocols result in different magnitudes and types of changes in movement variability (Nicol et al., 2006; Santamaria & Webster, 2010).

In addition to differences in test protocols, changes in performance output within a trial are common in both maximal and submaximal tasks (Figure 2.2). Tests of maximal or near-maximal effort often result in a clear decrement in performance output (Gollhofer, Komi, Fujitsuka, & Miyashita, 1987) as demonstrated by a significant decrease in maximal vertical jump height and vertical stiffness following repetitive maximal height jumps (Dal Pupo et al., 2013). Sprint time also significantly decreased in the final compared to the first trial by 0.59 s and 0.86 s for each sprinter (Trezise et al., 2011). Further, a significant decrease in mean vertical and medio-lateral GRF during the submaximal side cutting task was present post the fatiguing protocol (Cortes et al., 2014). Additionally, during the exhaustive submaximal run the authors reported an increasing number of speed adjustments occurring throughout the later stages of the trial in order to maintain the required average lap speed (Meardon et al., 2011).

As discussed (Section 2.2), movement variability is task specific (Bartlett et al., 2007; Stergiou & Decker, 2011), demonstrated by changes in the performance of a task effecting the magnitude of movement variability (Gates & Dingwell, 2011; James et al., 2000). Thus, within trial changes in performance output during continuous tasks may affect movement variability. Whether movement variability is affected by fatigue or differences in the task as fatigue progresses within a trial is difficult to discern. To be able to examine fatigue as a singular factor it is important to control the movement output parameters (Figure 2.2) such that any change in intrinsic characteristics may be attributed to fatigue only.

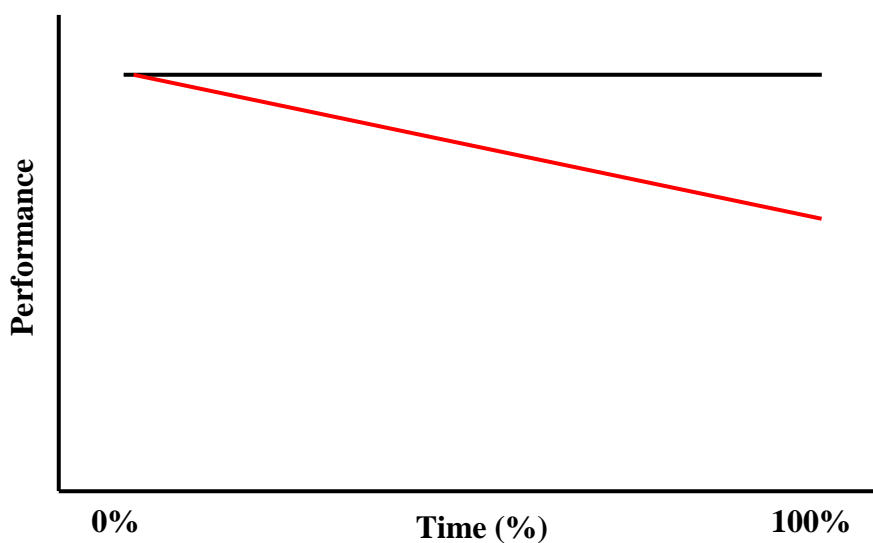


Figure 2.2 A theoretical schematic representation of performance output during a submaximal effort task completed to exhaustion. The red line indicates a decrease in task performance as fatigue progresses, which may confound the underlying neuromechanical changes. The black line indicates no change in performance output as fatigue progresses, which allows potential changes in neuromechanical characteristics to be better isolated to fatigue as a singular factor.

The definition of fatigue implies that the effects may be observed during an ongoing task. In a number of studies, dependent variables were only measured at the start and end of a fatiguing task (Cortes et al., 2014; Dal Pupo et al., 2013; Ferber & Pohl, 2011; Samaan et al., 2015; Trezise et al., 2011) and not monitored throughout the task. This is often because changes due to fatigue are assumed to be linear and thus the findings at the end of a fatiguing task will represent the overall trial. This has resulted in two common methods to induce fatigue, either to test movement before and after a separate fatiguing protocol or evaluate the start and end of a continuous task (Santamaria & Webster, 2010). However, changes due to fatigue have been found to be non-linear during a continuous run (Meardon et al., 2011). Thus, it has been difficult to determine whether the observed changes in movement were due to the inherent variability in repetitive tasks, the influence of motor learning due to repetition or fatigue. To determine what factor(s) may affect movement variability, monitoring throughout the duration of a continuous task is required.

Movement variability is important for the successful performance of a task, especially under changing conditions. With conflicting findings of the changes in movement variability with increasing levels of fatigue, greater insight into how fatigue affects movement variability can be gained from experiments involving a continuous task in which the performance output remains consistent as fatigue increases. Determining whether fatigue influences movement patterns and performance will extend the existing knowledge base.

2.4 Neuromuscular activity and fatigue

Maintaining a consistent performance output during a fatiguing dynamic task requires numerous compensatory mechanisms that occur at multiple levels of the musculoskeletal system. At a neuromuscular level, there have been shown to be modulation in both feedforward (central) and feedback (spinal-reflex) muscle activity for the regulation of task performance (Farley et al., 1998; Farley & Morgenroth, 1999; Hobara et al., 2007; Hortobágyi & DeVita, 2000). Further, these adaptations may occur as a change in amplitude and/or temporal characteristics within a single muscle or as a shift in relative activity between different muscles of the lower limb (Komi, 2000; Nicol et al., 2006).

Repetitive and fatiguing tasks that utilise the SSC such as running, jumping, hopping and sledge rebounds have resulted in a change in muscle function and reduced performance (Komi, 2000; Nicol et al., 2006). For example, there was demonstration of a decrease in amplitude of the electromyography (EMG) signal of the soleus (SOL), medial gastrocnemius (MG) and VL muscles at the end of continuous hopping on the ground (Kuitunen et al., 2002; Moritani et al., 1990) and on a sledge (Kuitunen et al., 2007; Morio et al., 2011). In contrast, there was a significant increase in SOL muscle activity towards the end of continuous submaximal height hopping (Oliver et al., 2014; Padua et al., 2006) and sledge rebounding (Regueme et al., 2005), and the short latency stretch-reflex occurred more frequently in the SOL muscle at the end of exhaustive double-leg hopping compared to the start (Maton & Pellec, 2001). Earlier SOL and tibialis anterior (TA) muscle onset and offset during sledge rebounds following a fatigue protocol demonstrated a temporal shift in muscle activity (Debenham, Travers, Gibson, Campbell, & Allison, 2015). Therefore,

previous research has demonstrated both increases and decrease in muscle activity during a fatiguing task.

It was reported that during these continuous and repetitive tasks there was a decreased height of hopping and displacement during the sledge rebound, demonstrating that there was a deterioration of performance following fatigue (Kuitunen et al., 2002; Kuitunen et al., 2007; Morio et al., 2011; Moritani et al., 1990). It is plausible that the reduced activity in the prime movers (triceps surae and quadriceps muscle groups) during hopping and sledge rebounding may lead to the deterioration in performance output. Consequently, the causality between reduced muscle activity and progression of fatigue remains difficult to interpret due to not controlling task performance of height of hopping or sledge rebound.

There are different compensatory strategies during a continuous task when a consistent performance is maintained (Komi, 2000; Nicol et al., 2006). Performance output during repetitive tasks such as hopping on the ground (Farley, Blickhan, Saito, & Taylor, 1991; Farley et al., 1998), on a sledge (Kuitunen et al., 2007; Morio et al., 2011) or overground running (Oliver et al. 2014) have been characterised using numerous variables including displacement of the COM, duration of flight and contact phases, changes in ground reaction force or leg stiffness. It is difficult to compare findings from different studies when some studies do not report measures of performance output (Bonnard et al., 1994) or due to the differences in the tasks and performance variables that were evaluated by each study. Thus it is not possible to determine with certainty whether performance was successfully maintained.

There are likely to be task specific differences between protocols that induce fatigue and what variables are characterised when evaluating the changes in neuromuscular activity. For example, running (Oliver et al., 2014) and squatting (Padua et al., 2006) to fatigue have been used as a task to induce fatigue when evaluating neuromuscular changes during double-leg hopping. Different fatiguing protocols are likely to influence the changes in neuromuscular activity measured during the hopping trial, particularly between compensatory muscle groups, and this may potentially confound responses between different muscle groups.

During repetitive tasks, neuromuscular adaptations may occur throughout. For example, over the duration of an exhaustive sledge rebound task there was an initial

decrease in feedforward activity followed by a significant increase until exhaustion (Regueme et al., 2005). Additionally, mechanical changes during fatigue have also been demonstrated with increased leg stiffness during the first half of the exhaustive sledge rebound task with no changes occurring thereafter (Morio et al., 2011). Whilst examining changes in neuromuscular characteristics during the performance of an exhaustive task it has been demonstrated that neuromuscular changes are non-linear (Morio et al., 2011; Regueme et al., 2005). Thus, any changes should be monitored throughout the performance of a dynamic repetitive loading lower limb task. Examining time dependent effects throughout the fatigue protocol may provide relevant information regarding what processes may affect neuromuscular characteristics during a continuous task to exhaustion.

2.5 Sex differences

Considerable demonstration of the differences between men and women in characteristics such as strength, power and agility have been shown (Thomas & French, 1985). Yet, it is not clear as to what differences there may be in the underlying neuromechanical characteristics or movement patterns particularly during a rapid and repeated dynamic task to exhaustion (Hunter, 2014). Women have been shown to have decreased lower limb movement variability compared to men during rapid loading dynamic tasks (Barrett et al., 2008; Breen, 2012; Maulder, 2011; Pollard et al., 2005). Findings when performing an un-anticipated cut task determined that women had significantly lower variability in thigh-leg rotation, thigh-leg abduction/adduction, knee flexion/extension-knee rotation and knee flexion/extension-hip rotation couplings compared to men during the initial 40% of the stance phase (Pollard et al., 2005). This finding was consistent with the landing phase of a land and cut task with a finding that thigh-leg rotation, hip rotation-knee abduction/adduction and knee rotation-knee abduction/adduction couplings were lower in women compared to men (Breen, Harrison, & Kenny, 2012). Variability of the tibia-knee rotation coupling was lower during the stance phase of an un-anticipated 180° turn for women compared to men (Maulder, 2011). During running, women had reduced variability of hip rotation, knee rotation and ankle flexion/extension angles at 12 km·h⁻¹ (Barrett et al., 2008) and lower ankle rotation at 8, 10 and 12 km·h⁻¹ (Barrett et al., 2008).

Women have demonstrated lower levels of movement variability than men during repeated lower limb loading tasks. Nevertheless, changes in lower limb movement variability during a fatiguing task in different sexes has only been explored in a homogenous sample group of men (Dal Pupo et al., 2013) or women (Samaan et al., 2015), or the aim did not involve investigating sex differences (Ferber & Pohl, 2011; Herb et al., 2014; Meardon et al., 2011; Miller et al., 2008). Therefore, it is unknown whether men and women adapt lower limb movement variability similarly during a continuous fatiguing dynamic task.

Neuromechanical differences between men and women suggest that women adopt a “quadriceps dominant” strategy during landing tasks (Griffin, Albohm, & Arendt, 2006; Shultz, Nguyen, Leonard, & Schmitz, 2009). This has been observed as greater rectus femoris muscle activity prior to landing (Nagano, Ida, Akai, & Fukubayashi, 2007) and vastus medialis activity during single-leg landing from a vertical jump (Urabe et al., 2005) in women compared to men. Following a fatiguing counter movement jump protocol men and women demonstrated similar values for mean integrated EMG (iEMG) of the quadriceps, hamstrings or gluteus maximus muscles during an unanticipated cut task (Iguchi et al., 2014). Further, both sexes demonstrated similar significant increases in gastrocnemius and SOL peak muscle activity during double-leg hopping following a fatiguing squat task (Padua et al., 2006). However, TA activity was significantly greater in women in the non-fatigued condition, but not in the fatigued condition (Padua et al., 2006). Women quadriceps (vastus medialis and rectus femoris muscles) activity was 45% higher than men during the non-fatigued and fatigued conditions, but was not significant due to large between-subject variance (Padua et al., 2006). Co-activation between the quadriceps and the hamstring muscles was significantly higher for women when non-fatigued and fatigued data were pooled (Padua et al., 2006).

A lack of significant post-hoc differences in neuromuscular activity between sexes may be due to small sample sizes (total participants = 26 (Gehring et al., 2009), 23 (Iguchi et al., 2014) and 21 (Padua et al., 2006)). It was also difficult to identify between which variables the effects lie as the data was pooled between fatigued and non-fatigued conditions (Gehring et al., 2009; Padua et al., 2006), sexes (Gehring et al., 2009; Iguchi et al., 2014; Padua et al., 2006) and/or different periods of muscle

activation (Padua et al., 2006). Further, fatiguing protocols that are different to the measurement task may induce task specific differences.

There remains a lack of understanding of differences between sexes in the underlying neuromuscular performance during dynamic fatiguing tasks. More mechanistic and well controlled studies are needed to better discern differences in neuromechanical characteristics between men and women during fatiguing dynamic tasks. Greater understanding of differences in neuromechanical characteristics during fatiguing tasks between sexes will promote more tailored and effective strategies to manage fatigue for both sexes.

2.6 Hopping as a model to investigate movement variability

Loading of the lower limb during locomotion tasks produces a complex interaction between a multi-segmental mechanical system and the muscles that act synergistically to control the motor task. Hopping is a movement most commonly performed by animals including birds and kangaroos (Cavagna & Kaneko, 1977), yet it is not a preferred gait pattern in humans. Nevertheless, hopping does allow performance of repetitive loading of the lower limbs, which mimics some of the characteristics of more common gait patterns such as walking and running (Blickhan, 1989; Farley et al., 1991; Lamontagne & Kennedy, 2013).

When aiming to examine functional changes in the human body during movement, the task investigated should be as natural as possible such as during daily or sporting activities (Nicol et al., 2006). Single- and double-leg hopping (Figure 2.3) when performed at a preferred frequency of ~2.2 Hz (132 hops per minute) is a demonstration of a multi-segmental coordinated task involving repeated loading of the lower limb (Farley et al., 1991; Hobara, Kobayashi, et al., 2013; Lamontagne & Kennedy, 2013; Nicol et al., 2006). Hopping in contrast to walking and running has been shown to have lower variability in performance output due to the constraints of the primary displacement of the COM being along the vertical plane only (Blickhan, 1989; Bryant, Crossley, Bartold, Hohmann, & Clark, 2011; Farley et al., 1991; Oliver & Smith, 2010; Pruyn et al., 2012). Hopping mimics other repetitive tasks such as running but can be controlled in terms of performance outcomes.

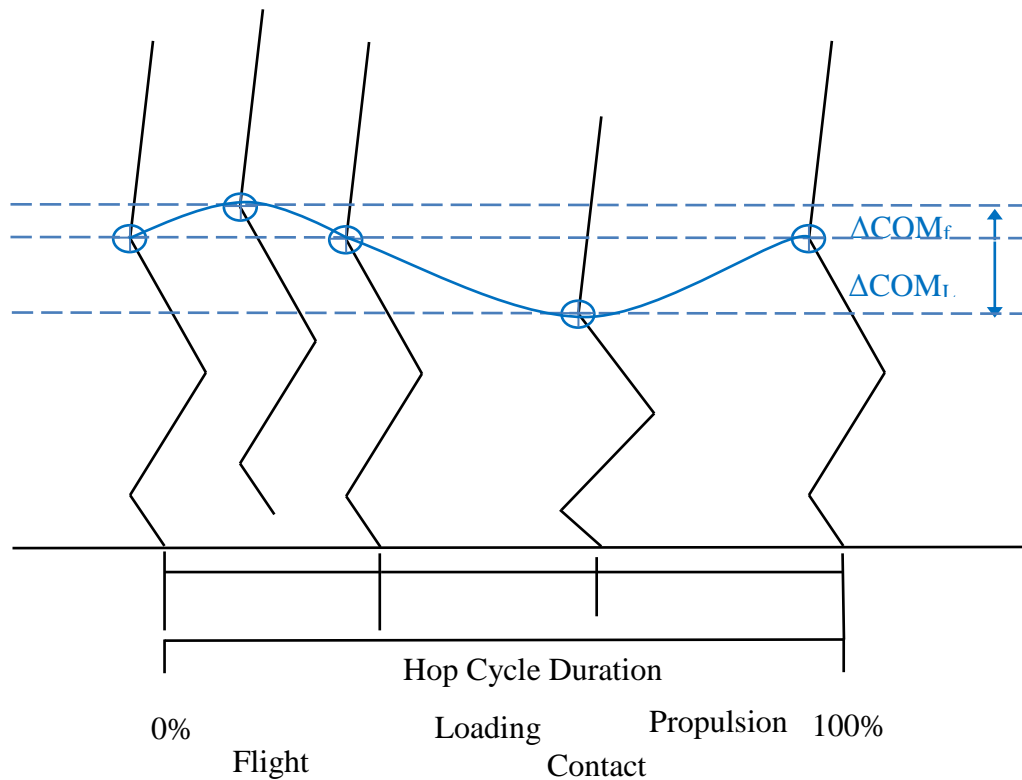


Figure 2.3 A schematic representation of on-the-spot hopping. A hop cycle consists of a complete flight phase and subsequent contact phase. The body's centre of mass (blue circle) maintains the motion of a harmonic sine wave throughout the hop cycle, with a positive and negative vertical displacement during the flight (ΔCOM_f) and loading (ΔCOM_L) phases, respectively.

Hopping on a sledge apparatus has been used to investigate difference in neuromechanics of the lower limb as it allows further control of the movement (Debenham et al., 2015; Furlong & Harrison, 2013; Komi, Horita, Kyröläinen, Takala, & Nicol, 1996; Kyröläinen et al., 1990; Morio et al., 2011; Morio et al., 2012; Regueme, Barthélemy, & Nicol, 2007; Regueme et al., 2005). This was achieved by limiting all motion to the lower limb only by fixing the participants trunk and the upper limbs on a sledge apparatus which the participant was then able to propel themselves by pushing off the force platform placed perpendicular to the rails of the sledge. By reducing the degrees of freedom available for hopping on the sledge apparatus compared to hopping on the ground, any changes introduced by alterations in trunk mechanics and spatial position of the participant relative to the force platform were minimised when compared to on-the-spot hopping. It should be noted that the sledge apparatus allows a greater focus on utilising the calf complex to

allow continuous hopping as there is a relatively greater constraint of the hip joint complex, thereby also constraining the amount of knee motion available (Furlong & Harrison, 2013; Morio et al., 2011). However, human locomotion is not constrained in this manner and on-the-spot hopping more closely mimics bouncing gaits such as walking and running, whilst still minimising confounding variations in performance compared to overground gait activities.

The focus of the current research was to determine the effect of fatigue on neuromechanical characteristics during repeated and rapid loading tasks to exhaustion. It was important to identify characteristics that were sensitive to changes in the level of fatigue and were not confounded due to differences in the performance output between consecutive gait cycles. Providing tactile feedback during on-the-spot single-leg hopping has been shown to allow participants to maintain performance output by a target height whilst maintaining the cadence (Gupta et al., 2016; Gupta, Mudie, & Clothier, 2014). Maintaining vertical displacement of the COM and spatial position of the participant has the benefit of allowing the investigation of the influence of fatigue on neuromechanical characteristics without the confounding factor of changes in performance. Therefore, hopping provides a simple and controlled model to investigate changes in underlying motor control and neuromuscular characteristics during fatiguing exercise.

2.7 Summary

A review of the effects of fatigue during dynamic tasks identified a variety of neuromechanical changes that occur, including adaptation in movement variability and muscle activity. Measures of movement variability can provide insight regarding adaptability of the neuromotor system. This can provide information on the state and health of the system during dynamic tasks that require multiple degrees of freedom. Whilst changes in movement variability are likely affected by subsequent changes in neuromuscular activity, or visa-versa, these characteristics are scarcely investigated in parallel during fatiguing tasks, and these strategies may differ between sexes.

Both time dependent and task dependent factors may influence changes in neuromechanical characteristics during a fatiguing task. To discriminate between these confounding factors, it is appropriate to evaluate a task that is sensitive to changes due to independent variables such as fatigue and sex, and have little change

due to other extraneous factors. Frequency and height controlled single-leg hopping provides a unique model to investigate adaptation strategies during a continuous fatiguing dynamic lower limb task. Based on the current understanding of the association between the health of the system and changes in movement variability, future investigations into how variability adapts under various conditions and constraints are required.


CHAPTER THREE


A comparison of vertical stiffness values calculated from different measures of centre of mass displacement in single-leg hopping

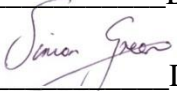
I, Kurt Mudie, hereby declare that I was the principal researcher of all work included in this chapter, including work published with multiple authors. A statement from the authors confirming the authorship and contribution of the PhD candidate to the published work is presented below.

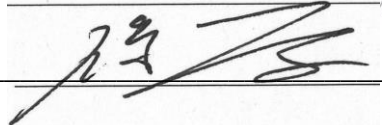
As co-authors of the paper ‘A comparison of vertical stiffness values calculated from different measures of centre of mass displacement in single-leg hopping’, we confirm that Kurt Mudie made the following contributions:

- Conception and design of the research questions
- Data collection, processing and analysis
- Statistical analyses and interpretation of findings
- Writing and review of the chapter/paper
- Corresponding author for communication with journal

Peter J Clothier:  Date: 6/01/17

Amitabh Gupta:  Date: 04/01/17

Simon Green:  Date: 16/01/2017

Hiroaki Hobara:  Date: Jan. 4 / 2017

The following publications have emanated from the study described in Chapter 3:

Mudie, K. L., Gupta, A., Green, S., Hobara, H., & Clothier, P. J. (2017). A comparison of vertical stiffness values calculated from different measures of centre of mass displacement in single-leg hopping. *Journal of Applied Biomechanics*. doi: 10.1123/jab.2016-0037. IF: 0.984 (Appendix A)

3.1 Introduction

Leg stiffness is often determined during human locomotion when describing the motion of the mass during landing tasks or bouncing gaits (Brughelli & Cronin, 2008; McMahon & Cheng, 1990). An increase in leg stiffness has been shown with a shorter contact time at a given hopping frequency (Farley et al., 1991; Hobara et al., 2007), or at a greater hopping frequency (Farley et al., 1991) and greater running velocity (Arampatzis, Bruggemann, & Metzler, 1999). Increasing leg stiffness may be a strategy which improves the utilisation of stored elastic energy within the musculoskeletal system in the lower extremity (Latash & Zatsiorsky, 1993). However, it is suggested that increased or decreased leg stiffness may also lead to a greater risk of injury due to a shift in the stress placed on the bony and soft tissues (Butler et al., 2003). Therefore, an accurate and consistent calculation of leg stiffness is required to allow a determination of a clinically significant change within or between subjects.

Leg and vertical stiffness (K_{vert}) are used synonymously during tasks where there is no horizontal translation, calculated as the quotient of the change in ground reaction force and displacement of the centre of mass (COM) (Blickhan, 1989). Although ground reaction force can be directly measured from a force platform, it is problematic to accurately measure the vertical displacement of the COM. Consequently, a number of methods have been developed to estimate the displacement of the COM to calculate K_{vert} (Butler et al., 2003; Hall, 2007; Hébert-Losier & Eriksson, 2014; Hobara et al., 2014; Ranavolo et al., 2008; Yang & Pai, 2014). The most common method used to calculate vertical COM displacement during stationary tasks involves double integration of the vertical acceleration measured from a force platform (Butler et al., 2003; Demirbüken, Yurdalan, Savelberg, & Meijer, 2009; Hébert-Losier & Eriksson, 2014; Hobara, Inoue, & Kanosue, 2013; Hobara et al., 2014; Hobara et al., 2010; Hobara et al., 2008; Hobara, Kobayashi, et al., 2013; Hobara, Kobayashi, Yoshida, & Mochimaru, 2015; Joseph, Bradshaw, Kemp, & Clark, 2014; Ranavolo et al., 2008). Less common methods using a segmental analysis and sacral marker have also been used to estimate COM displacement, but have been applied more often in tasks requiring horizontal translation (Eng & Winter, 1993; Gard, Miff, & Kuo, 2004; Iida & Yamamuro, 1987; Kerrigan, Viramontes, Corcoran, & LaRaia, 1995; Yang & Pai, 2014) rather than

vertical displacement (Ranavolo et al., 2008). A simplified method using flight and contact times to calculate vertical COM displacement has previously been proposed (Gupta et al., 2016; Hall, 2007) although not validated against multiple criterion measures.

During double-leg hopping at 2 and 2.4 Hz, Ranavolo et al. (2008) determined that motion capture of the sacral marker and the reconstructed pelvis methods produced significantly higher measures of vertical displacement of the COM compared to the segmental analysis and double integration methods. However, the authors (Ranavolo et al., 2008) did not explore the influence of these differences when computing K_{vert} . Further, Hébert-Losier and Eriksson (2014) determined that changing the integration constants used in the double integration method affected the measure of K_{vert} in double-leg hopping by changing the calculated initial velocity and displacement values. Lower K_{vert} values were also calculated when modelling the change in vertical ground reaction force (vGRF) as a sine-wave compared to the natural frequency of oscillation method and double integration method during double-leg hopping (Hobara et al., 2014). However, there is a significant difference in K_{vert} between single- and double-leg hopping tasks (Brauner et al., 2014; Hobara, Kobayashi, et al., 2013). The differences between computational methods for the calculation of K_{vert} may therefore be more or less pronounced in single-leg compared to double-leg hopping. Thus, as most athletic activities are performed unilaterally, differences in K_{vert} based on computations should be investigated in a single-leg loading task.

It remains unknown whether vertical displacement of the COM derived from different methods provides similar calculations of K_{vert} during single-leg hopping. The purpose of this study was to test the agreement between estimates of K_{vert} calculated using four different methods derived from either kinetic or kinematic-kinetic measures of the vertical displacement of the COM during single-leg hopping. It was hypothesised that significant differences in the calculation of K_{vert} and vertical displacement of the COM would be observed when using different methods.

3.2 Methods

3.2.1 Participants

Thirty-eight recreationally active males ($n = 20$) and females ($n = 18$) (mean \pm standard deviation (SD) 23.2 ± 2.2 years of age; 1.73 ± 0.06 m in height; 74.0 ± 10.4 kg in mass) volunteered to participate in this study. All participants were healthy and participated in exercise for between 1 to 4 hours per week. Participants had no past or current history of lower limb pathology, injury, pain, lower limb fracture or surgery within the six months prior to testing. Ethical approval (H1074) was granted by the University of Western Sydney Human Research Ethics Committee (Appendix F). Prior to testing, all participants read information describing the study (Appendix G), completed a pre-exercise health screen questionnaire (Appendix H) and provided written and informed consent (Appendix I).

3.2.2 Instrumentation

Kinetic and kinematic data were collected synchronously during single-leg, on-the-spot hopping from a multicomponent force plate (1500 Hz, Advanced Mechanical Technology, Inc., model BP400600-1000, Watertown, MA) and a three sensor Optotrak Certus System (150 Hz, Northern Digital Inc., Waterloo, Canada). All data were collected and time synchronised using First Principles (Version 1.2.4), and later processed using Visual 3D (C-Motion, Version 4, Germantown, MD). The audible tone of a metronome was set to the preferred hopping frequency for humans at 2.2 Hz (132 beats/minute) (Farley et al., 1991; Hobara, Kobayashi, et al., 2013).

A 16 marker cube was moved throughout the capture volume to spatially align all coordinate systems and a four marker probe used to define a right-handed laboratory coordinate system (Figure 3.1). The force platform was connected to a six-channel strain gauge signal conditioner and amplifier (Advanced Mechanical Technology, Inc., Gen 5, Watertown, MA) that were in turn connected to the multiplexer board by six Bayonet Neill-Concelman (BNC) connectors. The six channels represented the three dimensional force (F_x , F_y and F_z) and moment signals (M_x , M_y and M_z). A 50-pin ribbon cable (National Instruments, Australia) carried the analogue signals from the multiplexer board to an Optotrak Data Acquisition Unit II (ODAU II). The

ODAU II plugged into the Optotrak system control unit (SCU) allowing temporal synchronisation of all kinetic and kinematic data.

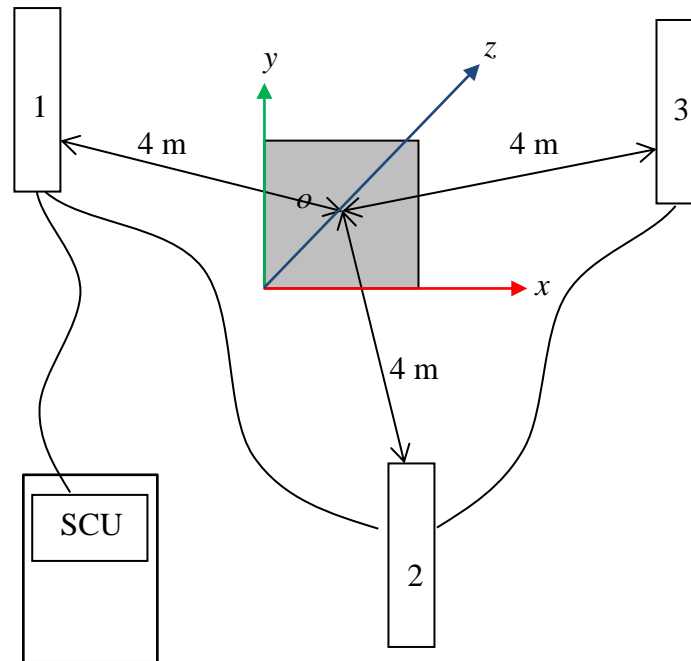


Figure 3.1 A schematic representation of the laboratory setup. The origin (o) was located at the back left corner of the force plate, the x -axis represented medio-lateral motion, y -axis anterior-posterior motion and the z -axis perpendicular to the x - y plane.

Custom-made three and four marker clusters were developed (Figure 3.2), conforming to previous guidelines (Cappozzo, Cappello, Croce, & Pensalfini, 1997). Custom files were created for individual clusters to define the fixed location of each marker relative to each other (NDI 6D Architect, Version 1.03.03). To minimise movement of the marker sets on the skin each marker cluster was tapered to ensure a smooth fit around the segment, adhered with double sided tape and secured by a Velcro strap (Figure 3.2).

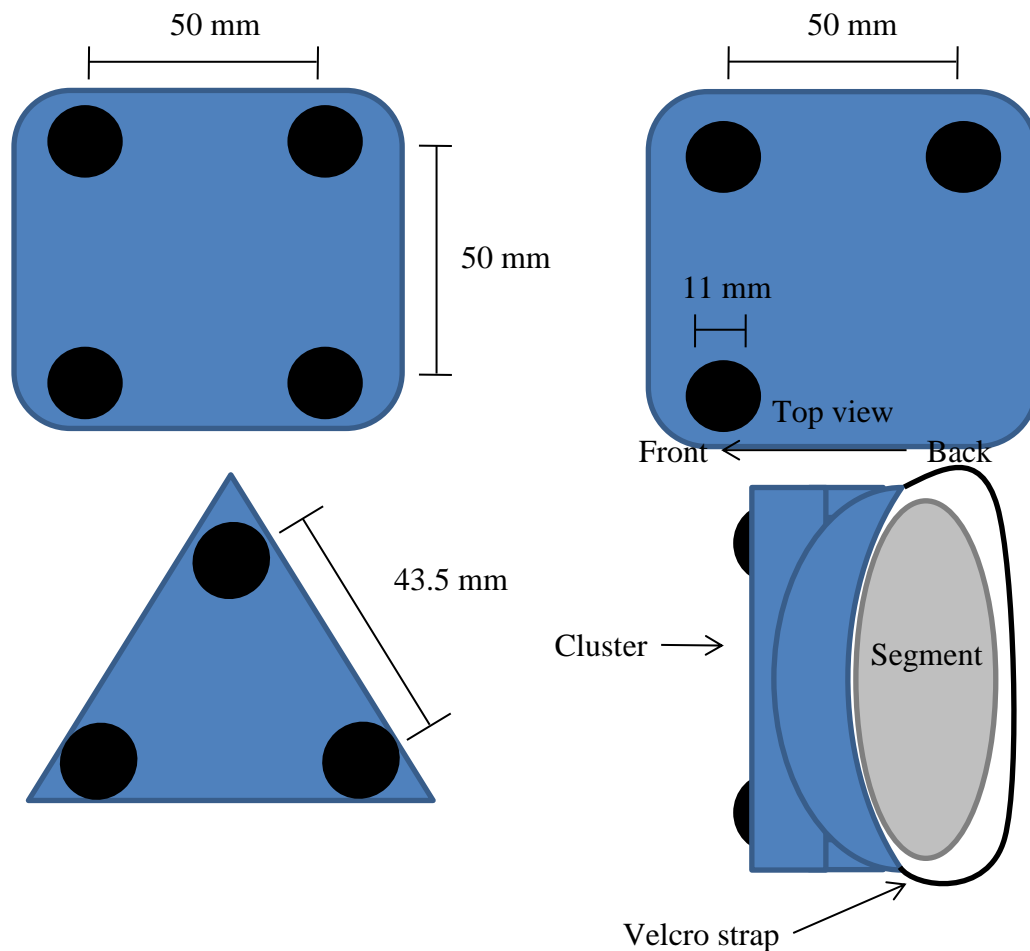


Figure 3.2 Custom-made three and four marker clusters. The back of each cluster was contoured to ensure a smooth fit around the segments (thigh and shank) and secured with a Velcro strap.

3.2.3 Participant preparation

Following measurement of each participant's height and body mass, a warm-up and a hopping familiarisation period was completed in accordance with that described by Hobara, Kobayashi, et al. (2013). This consisted of participants performing single-leg hopping to the metronome until they felt comfortable with the task (Hobara, Kobayashi, et al., 2013). Rest periods of 30 seconds were taken between hopping efforts to minimise the risk of fatigue. All trials were performed barefoot (Hobara, Inoue, Omuro, Muraoka, & Kanosue, 2011; Zuur et al., 2010) and on the participant's dominant leg, defined as their preferred kicking leg (Hobara, Inoue, et al., 2013; Padua et al., 2006). Each participant was instructed to keep their hands on their hips, hop on the ball of their foot, land on the beat of the metronome, minimise contact time and keep the knee of their non-hopping leg flexed at 90°.

3.2.3.1 *Kinematic model details*

A seven-segment model was used to model the trunk (G. Wu et al., 2005), pelvis (Bell, Pedersen, & Brand, 1990), thigh, shank (Ball, 2011), hindfoot, forefoot (W. L. Wu et al., 2000) and hallux (Stebbins, Harrington, Thompson, Zavatsky, & Theologis, 2006) of the hopping leg (Appendix J). Twenty one active markers were adhered over the segments, including a single four-marker cluster (sacral), three three-marker clusters (T8, distal lateral thigh and shank) and eight individual active markers (foot) (Cappozzo et al., 1997; Stagni, Fantozzi, Cappello, & Leardini, 2005). Fifteen calibration markers were digitised to identify the C7 and T8 spinous processes, suprasternal notch, xiphoid process, left and right acromioclavicular joints, left and right anterior superior iliac spines, left and right posterior superior iliac spines, greater trochanter, medial and lateral femoral epicondyles and medial and lateral malleoli of the dominant leg. Eight individual active markers (NDI, Optotrak, Canada) (7 mm diameter) were adhered to the posterior, medial and lateral aspects of the calcaneus, base and head of the first and fifth metatarsals and the medial aspect of the base of the proximal phalanx of the great toe (Figure 3.3). All foot markers were placed parallel with the ground using a custom made spirit level.

To minimise soft tissue movement artefact, marker clusters were placed away from areas of large muscle bulk and bony prominences (Cappozzo et al., 1997). Digitised calibration markers were chosen rather than physical markers as they have been reported to have greater reliability by minimising the influence of soft tissue artefact (Soutas-Little, 1996; Stebbins et al., 2006). To ensure reliability, only the primary investigator placed marker clusters and active markers on all participants. The outline of each marker cluster was marked on the participant's skin to confirm it remained in the same position throughout testing. All cables were secured with hypafix tape and the comfort of each participant was checked during warm-up and familiarisation efforts to ensure that there was no restriction of movement.

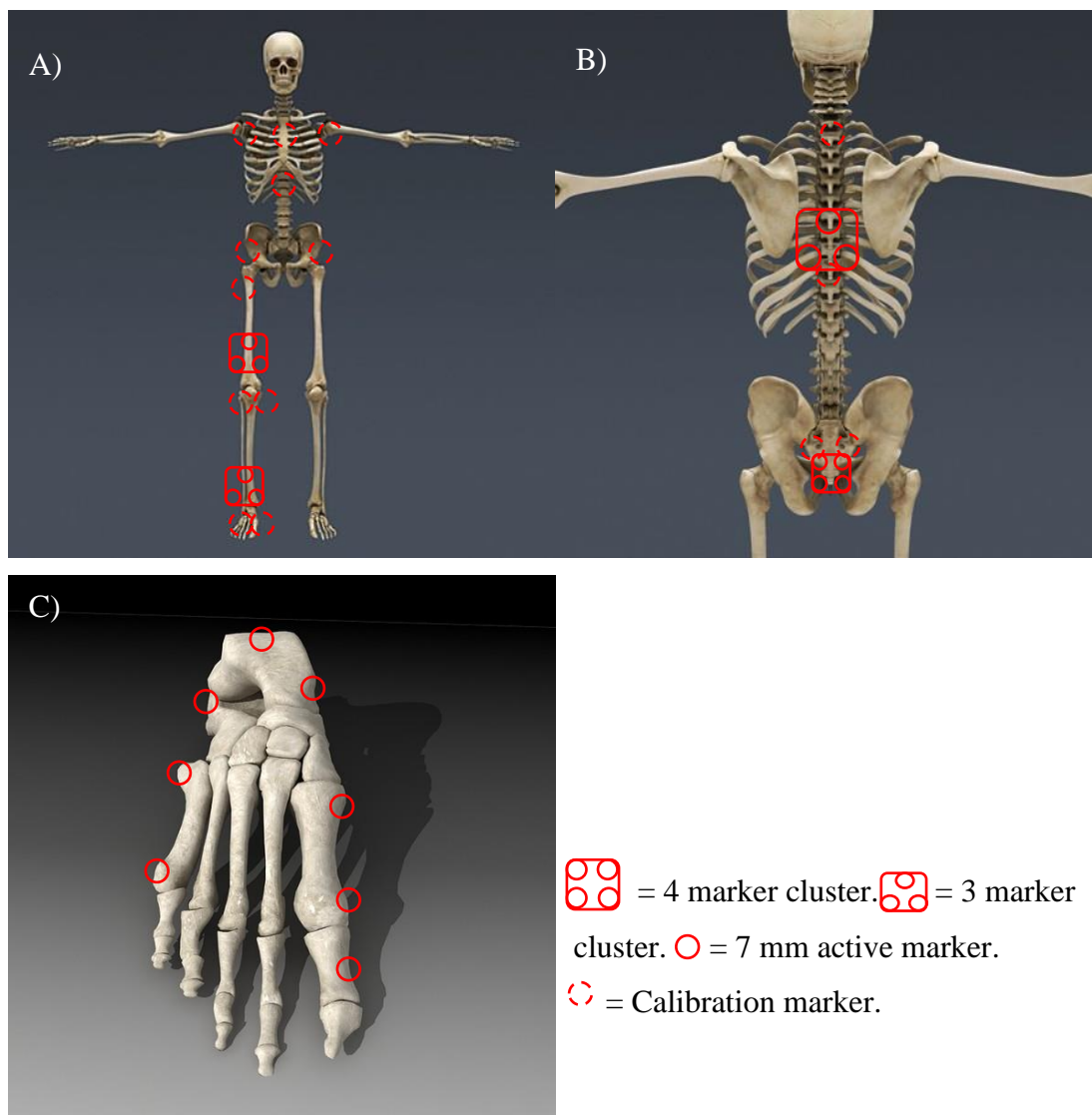


Figure 3.3 A) Anterior, B) posterior and B) superior view of active, calibration and cluster marker locations

The three segments represented by the foot model were the hindfoot (talus and calcaneus), forefoot (five metatarsals) and the toe (hallux), independent of one another (Carson, Harrington, Thompson, O'Connor, & Theologis, 2001; Stebbins et al., 2006) (Figure 3.4). The foot is commonly modelled as a single rigid segment (Farris & Sawicki, 2012; Risberg, Moksnes, Storevold, Holm, & Snyder-Mackler, 2009; Webster & Feller, 2012) that only allows net dorsiflexion/plantarflexion to be determined between the foot and shank (Carson et al., 2001). However in vivo measurements of foot kinematics during gait have demonstrated the complexity of foot and ankle motion by determining that there is movement at each of the articulations in the foot in multiple planes (Lundgren et al., 2008). Therefore, single

segment models of the foot have been shown to significantly overestimate ankle joint power during gait and may not represent the motion occurring in the foot region (MacWilliams, Cowley, & Nicholson, 2003). A three segment foot model, similar to that of the previously validated Oxford Foot Model (OFM) (Carson et al., 2001; Stebbins et al., 2006) and other two segment models (Doets, van Middelkoop, Houdijk, Nelissen, & Veeger, 2007; W. L. Wu et al., 2000) used during gait studies was considered appropriate based on the novelty of the hopping task. Hopping utilises a forefoot landing technique that has a similar technique and loading rate to running with a forefoot footfall pattern. Thus, it was assumed that there would be motion occurring at joints distal to the ankle during hopping similar to forefoot running. The three segment foot model allowed independent measurements of the joint angles between the shank – hindfoot (ankle), hindfoot – forefoot and forefoot – hallux segments (Lundgren et al., 2008).

In contrast to the OFM, the current model removed the marker between the second and third metatarsal heads and the calcaneal wand marker. The marker between the second and third metatarsal heads was replaced with a digitised marker at 53% of the distance between the first and fifth metatarsal, which has been shown to have no significant effect on measured joint angles and reliability (Stebbins et al., 2006). Further, removing the calcaneal wand marker and adding a single active marker on the posterior calcaneus eliminated a possible source of variability present with the risk of the wand marker being hit or displaced which has been reported to significantly improve the reliability of measured hindfoot rotation (Stebbins et al., 2006).

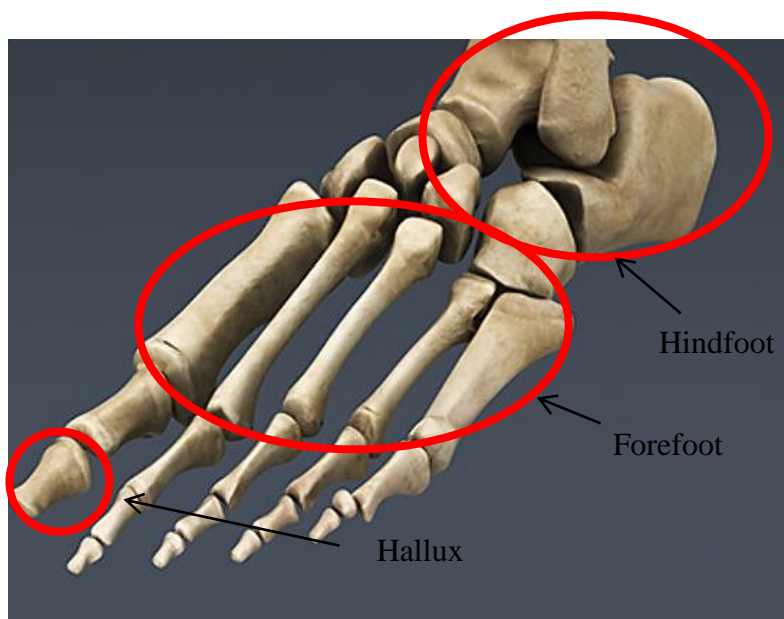


Figure 3.4 Schematic representation of the three segment foot model that consisted of the hindfoot (talus and calcaneus), forefoot (five metatarsals) and toe (hallux)

3.2.4 Testing protocol

A static calibration trial with the participant standing in the anatomical position was recorded. Each participant then completed a single effort of on-the-spot single-leg hopping at a cadence of 2.2 Hz for 15 s.

3.2.5 Data processing

All recorded trials were exported from First Principles software (Version 1.2.4) as C3D files for processing (Visual 3D, Version 4). A common scale factor and custom calibration matrix were used to convert raw analogue signals collected from the force plate to 3D force (N) and moment (N·mm) signals. The scale factor was set to manufacturer recommendations (Visual 3D, Version 4) and the calibration matrix calculated from a series of tests that compared collected analogue values (Visual 3D, Version 4) to calibrated force and moment signals (Biosoft, Version 2.3.1).

Kinematic marker recordings were interpolated using spline interpolation for up to a maximum gap of 10 frames. A third order polynomial equation was used and three frames before and after the gap used to calculate the coefficient of the equation. Recorded marker positions and force plate data were filtered using a fourth order

zero-lag bidirectional low-pass Butterworth filter with an 8 Hz (Bobbert & Richard Casius, 2011; Hobara et al., 2011; Hobara et al., 2008) and 50 Hz cut-off frequency (Gupta et al., 2016; Gupta et al., 2014), respectively.

To define model segments and joint centres the static trial was always processed before the movement trials. The defined segments included the trunk, thigh (femur), shank (tibia and fibula), hindfoot (calcaneus and talus), forefoot (five metatarsals) and hallux of the participant's hopping leg (Figure 3.5). Joint centres for the hip (pelvis – thigh) were calculated using the Bell regression equation (Bell, Brand, & Pedersen, 1989) and the midfoot (hindfoot – forefoot) as half the distance between the base of the first and fifth metatarsals (Morio, Lake, Gueguen, Rao, & Baly, 2009). Knee (thigh – shank) and ankle (shank – hindfoot) joint centres were calculated as half the distance between medial and lateral epicondyles and malleolus, respectively (G. Wu et al., 2002). Local coordinate systems for each segment were defined during the static trial consistent with the International Society of Biomechanics recommendations (Grood & Suntay, 1983; G. Wu et al., 2002; G. Wu et al., 2005) (Appendix K).

Joint axes were defined by the right-handed Cartesian local coordinate system for two adjacent body segments to determine the positive/negative direction of motion (G. Wu et al., 2002) (Figure 3.5). Motion of the segments were expressed as the orientation of the distal segment with reference to the neighbouring proximal segment (Doets et al., 2007; Farris & Sawicki, 2012; Stebbins et al., 2006; G. Wu et al., 2002). Medio-lateral (y) and superior-inferior (z) axes were mirrored in the left limbs to ensure consistency between sides regarding positive/negative motion about a joint axis (Grood & Suntay, 1983).

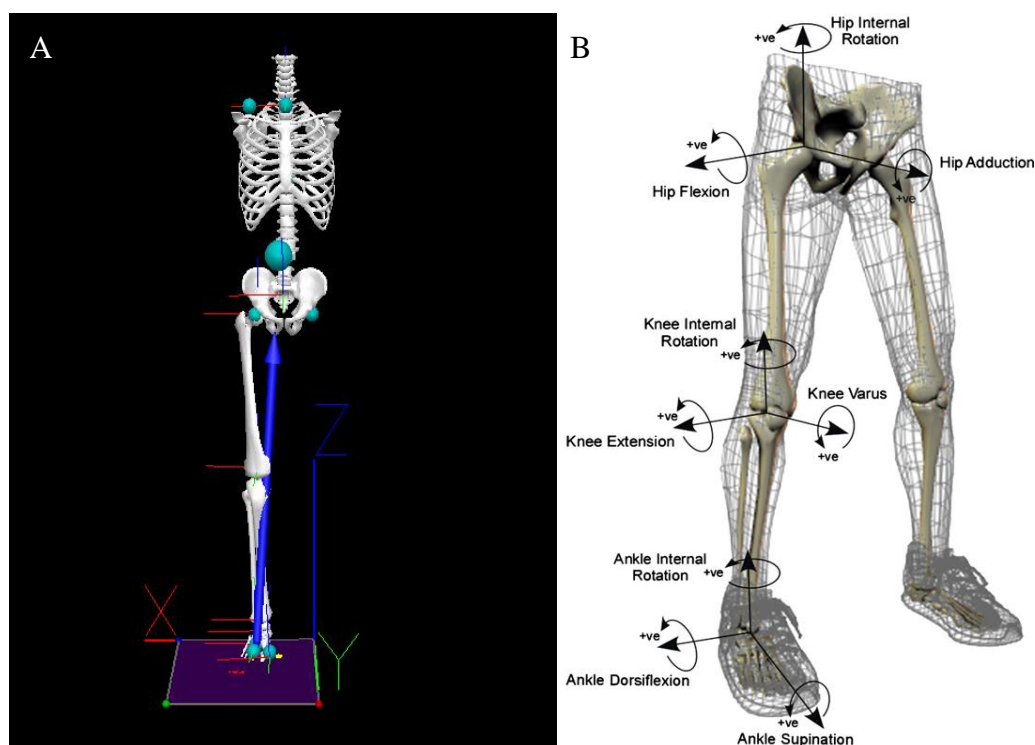


Figure 3.5 A) Reconstructed model segments and joint rotation axes in Visual 3D. B) Joint coordinate system and direction of 3D joint rotations for the lower limb joints, adapted from (McLean, Huang, & van den Bogert, 2008).

3.2.6 Data analysis

A single hop cycle was defined as a complete flight phase and the subsequent contact phase (Figure 3.6). The contact phase included the loading (initial contact to peak vGRF) and propulsive (peak vGRF to toe-off) phase. Temporal events including initial contact and toe off were determined for each hop cycle from the vGRF trace as the first and last point ≥ 10 N (Lloyd, Oliver, Hughes, & Williams, 2009), respectively. Peak vGRF was defined as the maximal vGRF value during the contact phase.

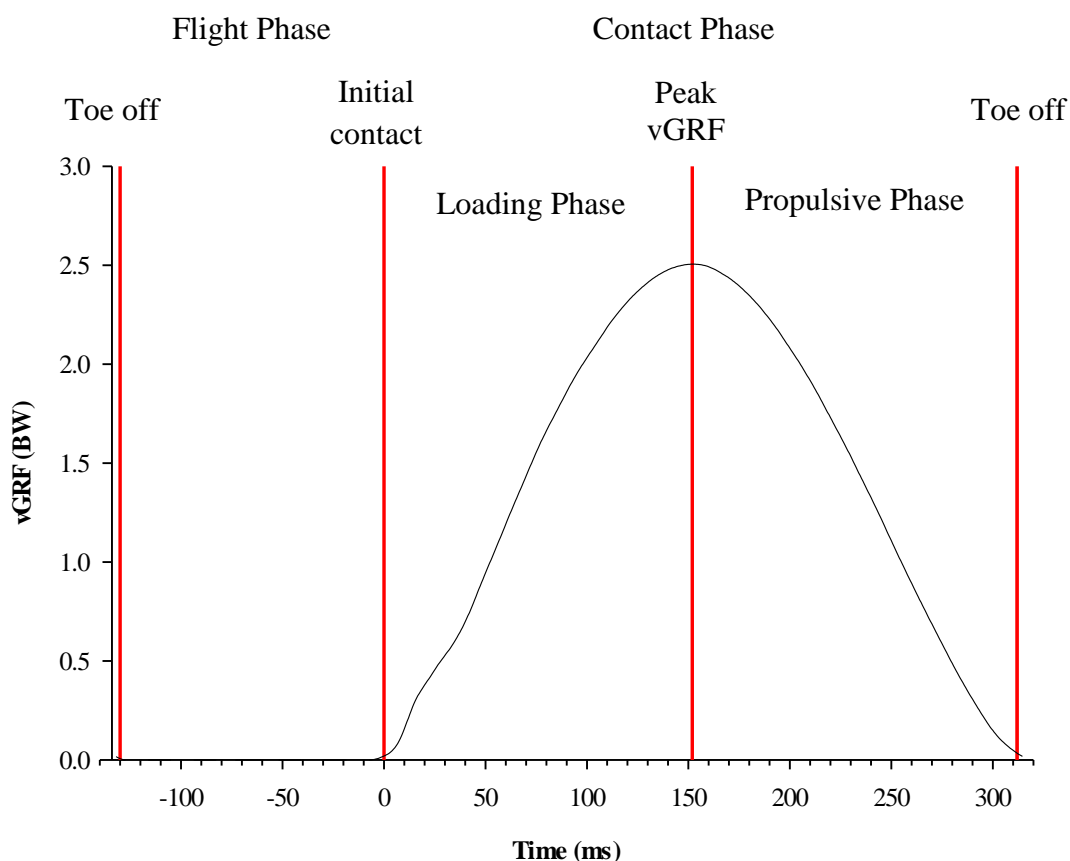


Figure 3.6 Temporal events defined during a single hop cycle.

The first twenty hop cycles were extracted from the trial data for each participant ($n = 760$). Seven hop cycles from five separate participants were incomplete or contained missing kinematic data and were removed from the analysis. A final total of 753 complete hop cycles were analysed from all participants. Dependent variables calculated for all hop cycles included K_{vert} and vertical displacement of the COM during flight (ΔCOM_f) and loading phases (ΔCOM_L). Each dependent variable was calculated using the sacral marker cluster, segmental analysis, double integration and law of falling bodies methods.

Sacral marker cluster method. The vertical displacement of the sacral marker cluster was used to estimate ΔCOM_f and ΔCOM_L (Ranavolo et al., 2008; Yang & Pai, 2014).

Segment analysis method. The vertical displacement of the COM was calculated as the weighted average of the seven modelled upper and lower limb segments to estimate ΔCOM_f and ΔCOM_L (Ranavolo et al., 2008; Yang & Pai, 2014). The

instantaneous position of the COM was estimated using the location and masses of each of the seven modelled segments (Equation 3.1) (Saini, Kerrigan, Thirunarayan, & Duff-Raffaele, 1998):

$$\text{COM}(t) = \frac{1}{M} \sum_{i=1}^n m_i y_i(t) \quad (3.1)$$

where n was the total number of segments, including the trunk, pelvis, thigh, shank and foot (inclusive of the hindfoot, forefoot and hallux), m_i was the mass of a segment, $y_i(t)$ was the instantaneous location of the COM of a segment and M was the sum of the masses of the segments. The segment masses and inertial properties were derived from Dempster (1955) (Dempster, 1955) and Hanavan and Ernest (1964) (Dempster, 1955), respectively.

Double integration method. Vertical displacement of the COM was estimated by integrating the vertical acceleration of the COM twice to calculate values for ΔCOM_f and ΔCOM_L (Hébert-Losier & Eriksson, 2014; Hobara, Kobayashi, et al., 2013). Vertical acceleration of the COM was first obtained from the vGRF curve at each frame, $1 \leq i \leq n$ as (Equation 3.2):

$$a_i = \frac{1}{m} f_i - g \quad (3.2)$$

where i was each frame that was less than or equal to n , the total number of frames, m was body mass and g was the acceleration due to gravity ($9.81 \text{ m}\cdot\text{s}^{-2}$).

Vertical velocity was then obtained by integrating the vertical acceleration at each frame as (Equation 3.3a and b):

$$v_i = v_o + a_i \cdot \Delta t \quad (3.3a)$$

$$v_{i+1} = v_i + a_{i+1} \cdot \Delta t \quad (\text{for } 1 \leq i \leq n) \quad (3.3b)$$

where Δt is the change in time between frames ($1/1500 \text{ s}$) and the first integration constant v_o was the velocity of the COM at initial contact as (Equation 3.4):

$$v_o = -\frac{1}{2} \cdot g \cdot t_f \quad (3.4)$$

where t_f was the duration of the flight phase.

Vertical displacement was obtained by integrating the vertical velocity at each frame as (Equation 3.5):

$$s_{i+1} = s_i + v_{i+1} \cdot \Delta t \quad (3.5)$$

where the second integration constant (vertical displacement of the COM at initial contact) was considered zero.

Law of falling bodies method. The vertical displacement of the COM was estimated by the law of falling bodies (Gupta et al., 2016; Hall, 2007) that assumes a parabolic displacement-time curve during the flight and contact phase and there was a change in direction of the COM at mid-flight and peak vGRF when $v = 0 \text{ m}\cdot\text{s}^{-1}$.

The vertical displacement of the COM during the flight phase was calculated by (Equation 3.6):

$$\Delta\text{COM}_f = \frac{1}{2} \cdot g \cdot \left(\frac{t_f}{2}\right)^2 \quad (3.6)$$

The velocity of the COM at initial contact was determined by (Equation 3.7):

$$v_o = \sqrt{2 \cdot g \cdot \Delta\text{COM}_f} \quad (3.7)$$

The vertical displacement of the COM during the loading phase was then calculated by (Equation 3.8):

$$\Delta\text{COM}_L = \frac{v_o + v_f}{2} \cdot t_l \quad (3.8)$$

where v_f was the velocity of the COM at peak vGRF ($0 \text{ m}\cdot\text{s}^{-1}$ due to a change in direction assumed at peak vGRF) and t_l was the duration of the loading phase.

Vertical stiffness was then calculated for each method as the quotient of force and displacement (Equation 3.9) (Butler et al., 2003):

$$K_{\text{vert}} = \frac{\text{peak vGRF}}{\Delta\text{COM}_L} \quad (3.9)$$

3.2.7 Statistical analyses

To quantify the agreement between methods Bland-Altman plots were created for each method pair for all three dependent variables (Bland & Altman, 1986). Specifically, the difference between the methods was plotted against their mean for each hop cycle (SigmaPlot, Version 12.5). The bias was determined as the mean difference across all hop cycles and the 95% limits of agreement (LoA) calculated as the mean difference (bias) $\pm 1.96 \times$ SD of the difference (Bland & Altman, 1999). To quantify trends of the bias over a range of values, the slope of a regression line was calculated for each Bland-Altman plot. The coefficient of variation for each method was calculated as the SD normalised to the mean and converted to a percentage (%) to determine the variance by each method.

3.3 Results

Mean, minimum and maximum values over all hop cycles were calculated for hopping frequency and peak vGRF (Table 3.1). Mean K_{vert} was greatest using the law of falling bodies method, followed by the double integration, segmental analysis then sacral marker cluster methods (Table 3.2). For the calculation of K_{vert} across individual hop cycles, Bland-Altman plots showed that the double integration and segmental analysis methods had the greatest agreement with a relatively small bias ($0.93 \text{ kN}\cdot\text{m}^{-1}$) and 95% LoA (-1.89 to $3.75 \text{ kN}\cdot\text{m}^{-1}$) (Figure 3.7). Further, the segmental analysis and double integration methods had the lowest coefficient of variation (Table 3.3) and slope of the regression line. In contrast, a greater bias, 95% LoA, coefficient of variation (Table 3.3) and slope of the regression line were calculated between the sacral marker cluster and the double integration ($-3.25 \text{ kN}\cdot\text{m}^{-1}$, -6.62 to $0.11 \text{ kN}\cdot\text{m}^{-1}$) and segmental analysis methods ($-2.32 \text{ kN}\cdot\text{m}^{-1}$, -4.40 to $-0.25 \text{ kN}\cdot\text{m}^{-1}$) (Figure 3.7). The law of falling bodies method revealed a relatively large bias and slope of the regression line when compared to the segmental analysis ($18.19 \text{ kN}\cdot\text{m}^{-1}$), double integration ($17.26 \text{ kN}\cdot\text{m}^{-1}$) and sacral marker cluster ($20.52 \text{ kN}\cdot\text{m}^{-1}$) methods (Figure 3.7).

Table 3.1 Hop frequency (mean (SD), minimum and maximum) and peak vGRF for all 753 hop cycles.

	Mean (SD)	Range
Hop Frequency (Hz)	2.22 (0.09)	2.00 – 2.83
Peak vGRF (kN)	1.77 (0.34)	1.11 – 2.62

Table 3.2 Vertical stiffness (K_{vert}) (mean (SD)) and vertical displacement of the centre of mass during loading (ΔCOM_L) and flight (ΔCOM_f) phases calculated from the sacral marker cluster, segmental analysis, double integration and law of falling bodies methods.

Method	K_{vert} (kN.m ⁻¹)	ΔCOM_L (cm)	ΔCOM_f (cm)
Sacral marker cluster	15.33 (3.27)	11.62 (1.13)	2.47 (1.15)
Segmental analysis	17.65 (3.58)	10.07 (0.96)	2.01 (0.98)
Double integration	18.58 (3.84)	9.59 (1.07)	2.52 (0.99)
Law of falling bodies	35.85 (7.66)	5.01 (0.76)	2.54 (0.99)

Table 3.3 Coefficient of variation (%) of the sacral marker cluster, segmental analysis, double integration and law of falling bodies methods for each dependent variable.

Coefficient of Variation (%)	Sacral marker cluster	Segmental analysis	Double integration	Law of falling bodies
Vertical Stiffness	21.33	20.27	20.64	21.38
Loading Height	9.70	9.54	11.21	15.17
Flight Height	46.45	49.10	39.17	39.01

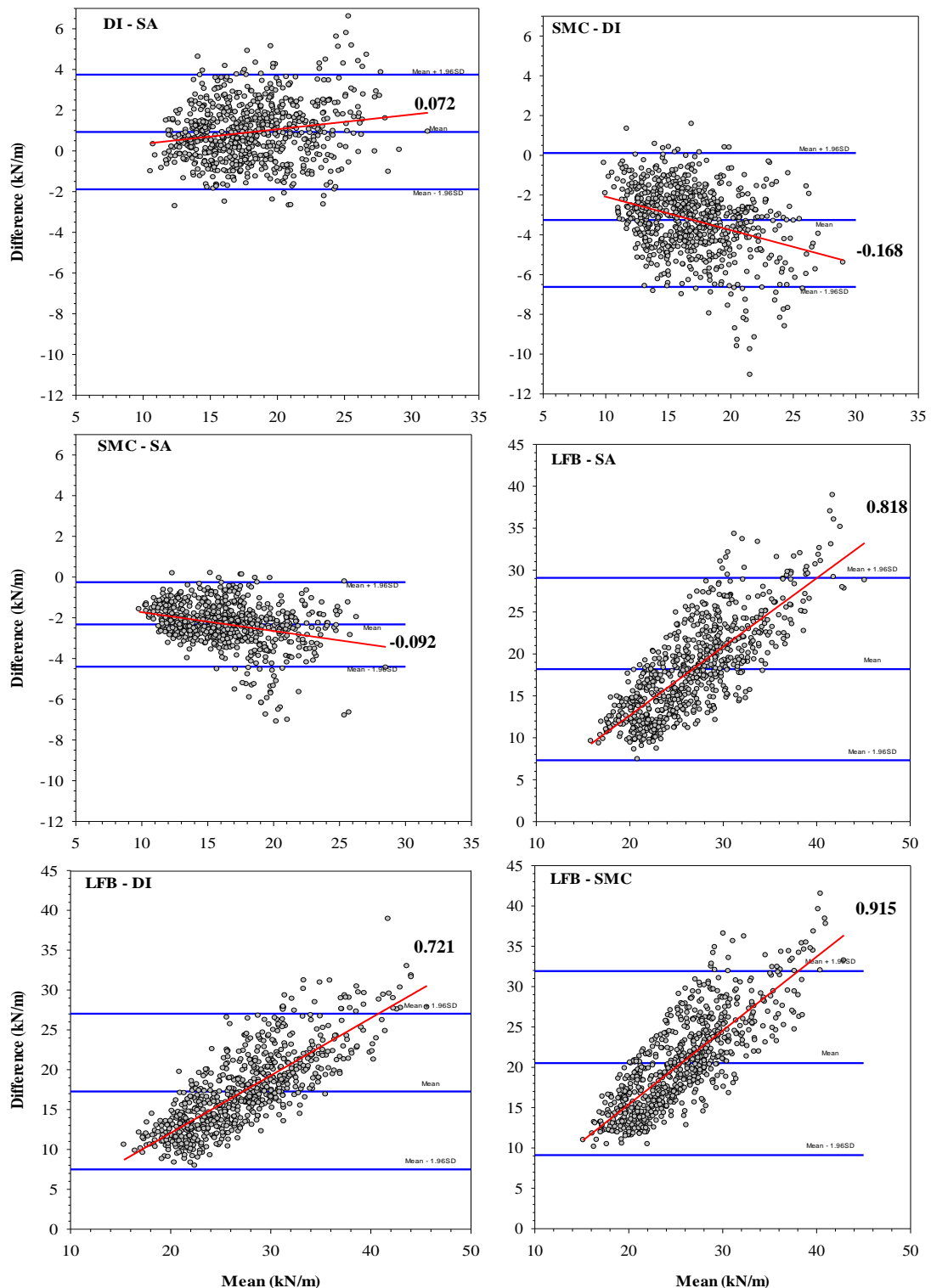


Figure 3.7 Bland-Altman plots for the comparison of vertical stiffness between the sacral marker cluster (SMC), segmental analysis (SA), double integration (DI) and law of falling bodies (LFB) methods. The difference (bias) is represented by the inner blue horizontal line, 95% limits of agreement the outer blue lines, regression line the red line and the magnitude of the regression slope labelled to the right.

Mean ΔCOM_L was lowest using the law of falling bodies method, followed by the double integration, segmental analysis then sacral marker cluster methods (Table 3.2). For the calculation of ΔCOM_L across individual hop cycles, the double integration and segmental analysis methods demonstrated the greatest agreement (bias and 95% LoA) (-0.48 cm and -1.96 to 1.00 cm) (Figure 3.8). Further, the segmental analysis and double integration methods had the lowest coefficient of variation (Table 3.3) and slope of the regression line. Bland-Altman plots revealed a relatively large bias but similar 95% LoA between the sacral marker cluster and the segmental analysis (1.55 cm and 0.26 to 2.84 cm) and double integration methods (2.03 cm and 0.05 to 4.02 cm). The law of falling bodies method revealed a relatively large bias, coefficient of variation (Table 3.3) and slope of the regression line when compared to the segmental analysis (-5.06 cm), double integration (-4.58 cm) and sacral marker cluster (-6.61 cm) methods (Figure 3.8).

Mean ΔCOM_f was greatest using the law of falling bodies method, followed by the double integration, sacral marker cluster then segmental analysis methods (Table 3.2). For the calculation of ΔCOM_f across individual hop cycles, Bland-Altman plots showed the greatest agreement (bias and 95% LoA) between the law of falling bodies and double integration method pair (0.02 cm and -0.01 to 0.03 cm) (Figure 3.9). Further, the law of falling bodies and double integration methods had the lowest coefficient of variation (Table 3.3) and no regression slope. A small bias and 95% LoA were also revealed between the sacral marker cluster and double integration method pair (-0.05 cm and -1.13 to 1.02 cm) and the sacral marker cluster and law of falling bodies method pair (0.08 cm and -1.00 to 1.16 cm). The greatest bias was between the segmental analysis and the sacral marker cluster (0.46 cm), law of falling bodies (0.54 cm) and double integration (0.52 cm) method pairs (Figure 3.9). There was a relatively large regression slope between the sacral marker cluster and all methods and a small regression slope between the segmental analysis and double integration method.

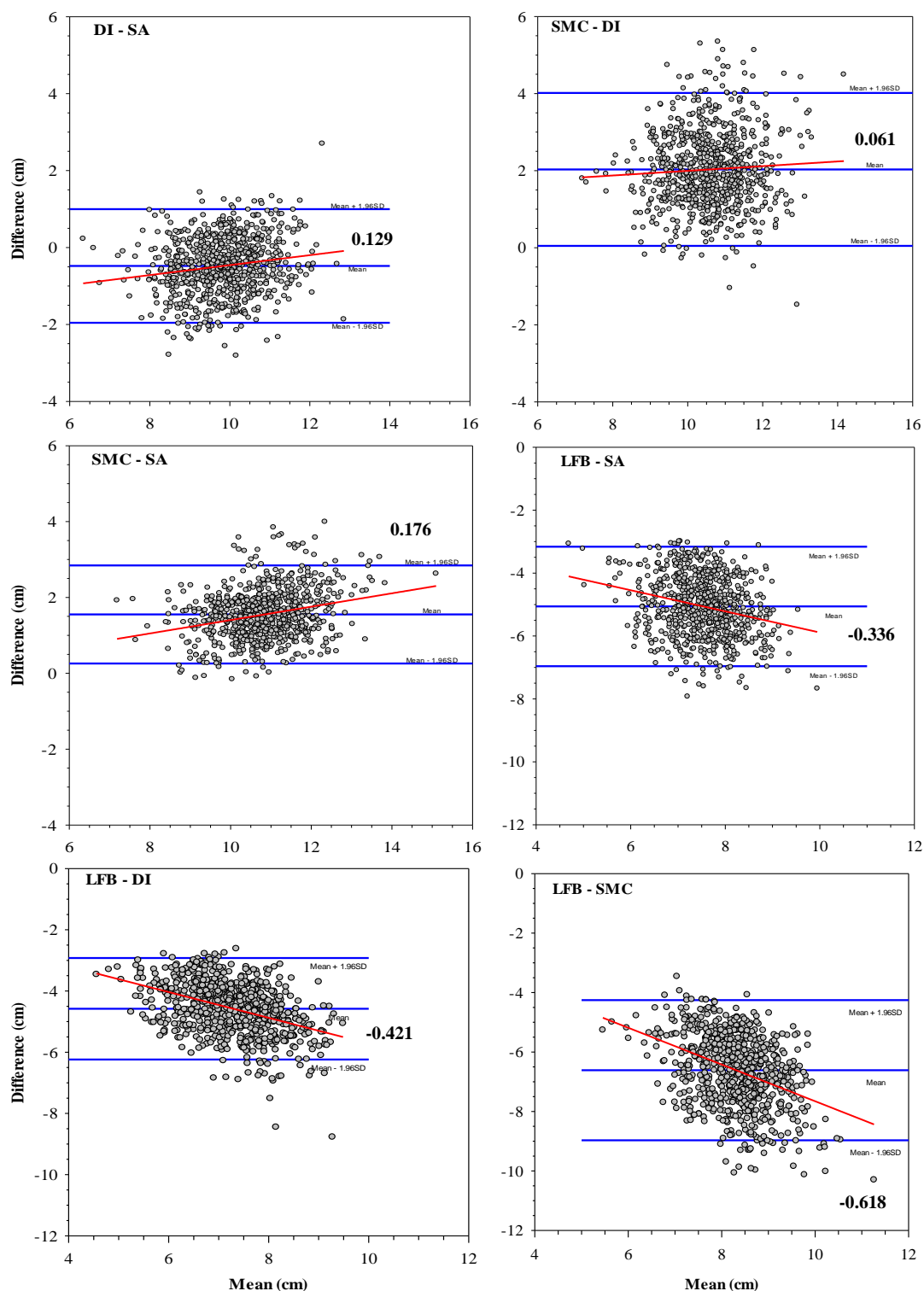


Figure 3.8 Bland-Altman plots for the comparison of the vertical displacement of the COM during the loading phase between the sacral marker cluster (SMC), segmental analysis (SA), double integration (DI) and law of falling bodies (LFB) methods. The difference (bias) is represented by the inner blue horizontal line, 95% limits of agreement the outer blue lines, regression line the red line and the magnitude of the regression slope labelled to the right.

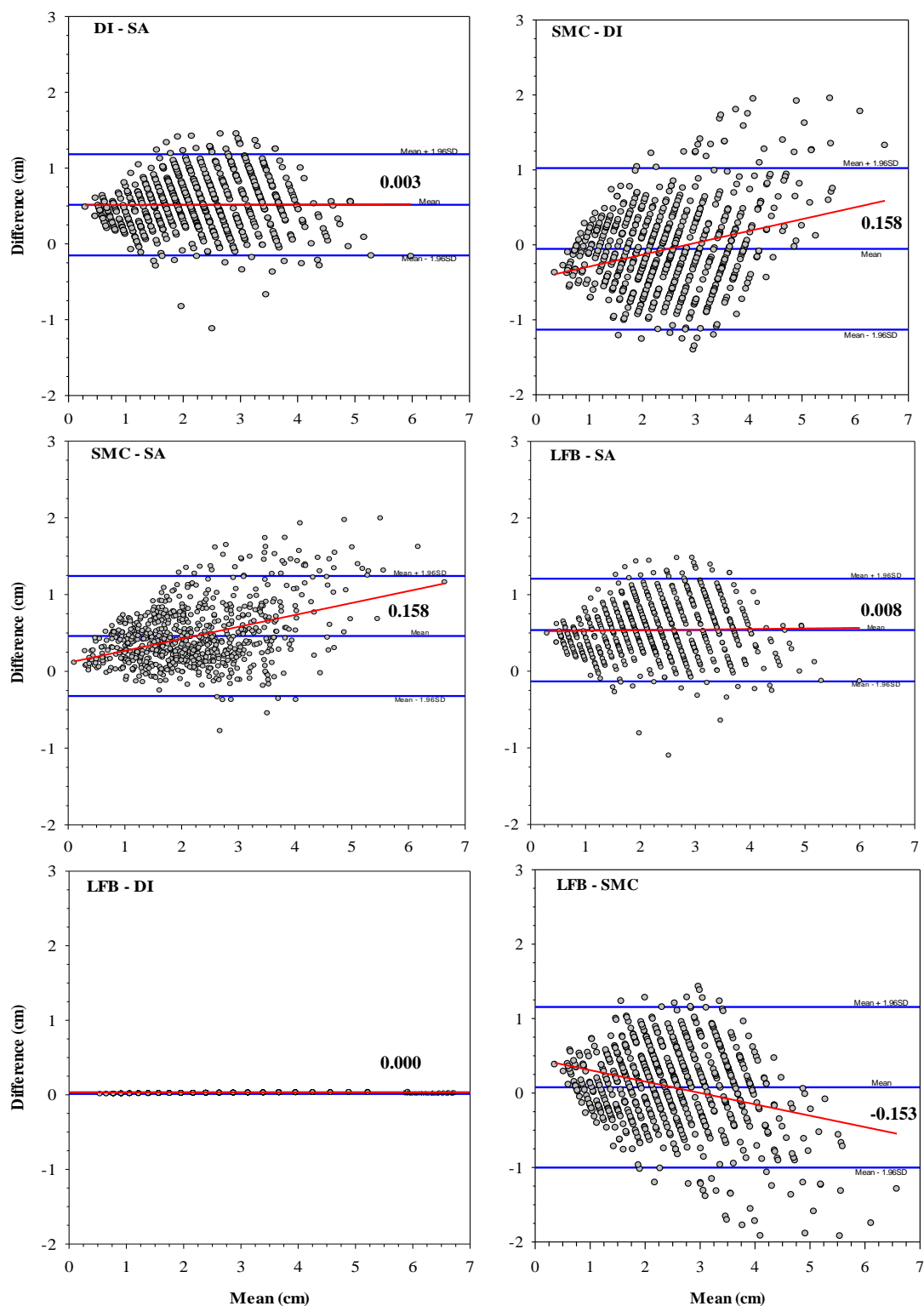


Figure 3.9 Bland-Altman plots for the comparison of the vertical displacement of the COM during the flight phase between the sacral marker cluster (SMC), segmental analysis (SA), double integration (DI) and law of falling bodies (LFB) methods. The difference (bias) is represented by the inner blue horizontal line, 95% limits of agreement the outer blue lines, regression line the red line and the magnitude of the regression slope labelled to the right.

3.4 Discussion

The purpose of this study was to assess the agreement between vertical stiffness calculated from two kinetic and two kinematic-kinetic methods of estimating the vertical displacement of the COM during single-leg hopping. The main finding of this investigation was that K_{vert} calculated using the segmental analysis and double integration methods were very similar, with the lowest coefficient of variation during a single-leg loading task. The authors' propose the segmental analysis method can be considered to represent the gold standard for the calculation of K_{vert} and vertical displacement of the COM during single-leg, on-the-spot hopping. Further, the current results support the use of either the segmental analysis or double integration methods to calculate K_{vert} during single-leg hopping.

When comparing the segmental analysis and double integration methods for the calculation of K_{vert} , Bland-Altman plots revealed only a small bias ($< 1 \text{ kN}\cdot\text{m}^{-1}$) over a range of stiffness values from 10.58 to 31.12 $\text{kN}\cdot\text{m}^{-1}$. Specifically, Bland-Altman plots revealed small 95% LoA and a small proportional systematic error as mean K_{vert} increased between the segmental analysis and double integration methods. Further, mean K_{vert} from the segmental analysis and double integration methods were most comparable to previous research (Brauner et al., 2014; Hobara, Kobayashi, et al., 2013; Joseph et al., 2014) with lower variance than the sacral marker cluster and law of falling bodies methods.

Assumptions of the segmental analysis method are limited to the segment properties and complexity of the model (number of segments modelled). Modelled segments are assumed to be rigid with dimensions calculated from anthropometric measurements of cadaver specimens (Gard et al., 2004; Ranavolo et al., 2008; Yang & Pai, 2014). Since the calculation of the COM position is based on the weighted average and position of the modelled segments, any movement of non-modelled segments that would theoretically change COM position cannot be detected. For the current study, since only one leg and the trunk were modelled there was a shift in the position of the COM towards the ipsilateral side. Further, any movement of the arms or contralateral non-weightbearing leg would not be detected. However, the testing protocol aimed to nullify the potential error by minimising movement of the upper limbs and contralateral non-weightbearing leg. Future researchers investigating

single-leg loading tasks may choose to only model the side of interest for ease of testing and to save time during the experimental set-up, eliminating unnecessary data.

The assumptions of the double integration method relate to the calculations which may smooth the displacement signal, potentially explaining the lower COM displacement values compared to the segmental analysis method (Ranavolo et al., 2008). In addition, determination of the velocity integration constant assumes symmetry of the hop cycle. Due to the asymmetry of the hop cycle and potential differences within and between subjects these assumptions are incorrect and will impact the calculated K_{vert} . As the value of the initial velocity integration constant has been shown to produce approximately 14% variance of the calculated K_{vert} values during double-leg hopping (Hébert-Losier & Eriksson, 2014), the chosen method should be explained or referenced in each instance. Accurate calculation of K_{vert} from the double integration method also requires motion of the body to mimic a simple spring-mass model, which may not represent the motion of hopping at lower frequencies (i.e. < 2 Hz).

The use of a double integration method for calculation of K_{vert} has the benefit of only requiring a force platform. The capture of 3D motion has the additional benefit of providing the investigator with positional data on the COM that is determined from the weighted average of the modelled segments. In contrast to the double integration method, the segmental analysis method is not limited by the asymmetry of the hop cycle or changes in COM position during the hop cycle. Thus, the segmental analysis method is sensitive to variations in the task and subject, by not modelling the subject as a simple spring-mass and likely contributing to the lowest variance for the calculation of K_{vert} and ΔCOM_L . Further, due to the built-in calculations within Visual3D, the simplistic nature of the segmental analysis method and the increased availability and use of 3D motion capture equipment within human movement laboratories, there is a justifiable reason to use the segmental analysis method as the gold standard for the calculation of K_{vert} during single-leg, on-the-spot hopping. Despite the assumptions of the double integration method it maintained a good agreement with the more sophisticated segmental analysis method for calculating K_{vert} and should be considered appropriate if there is only force plate data available.

Calculation of K_{vert} was consistently lower using the sacral marker cluster method due to higher values of ΔCOM_L . The positive bias observed on Bland-Altman plots highlight a proportional systematic error between the sacral marker cluster and the other three methods when calculating K_{vert} , ΔCOM_f and ΔCOM_L . In contrast to the segmental analysis method, the sacral marker cluster method assumes a stationary COM position and cannot detect changes in COM position when other body segments move. Further, when using a direct measure as an estimate of the COM position confounding factors such as pelvic tilt and movement of the sacral rigid body on the skin during the hopping task may overestimate movement of COM position. These errors may contribute to the relative higher values of the vertical displacement of the COM and proportional systematic error.

Although, there was a smaller bias between the sacral marker cluster and other methods when the measure of ΔCOM_f and ΔCOM_L were relatively small in magnitude, this is most likely due to the minimisation of the errors when there is relatively less displacement and movement of the body's segments. Therefore, the sacral marker cluster may be an appropriate method for the calculation of vertical displacement of the COM for single-leg tasks with relatively less displacement, such as when hopping at higher frequencies (Hobara et al., 2010; Ranavolo et al., 2008). In contrast, there was a greater bias between the sacral marker cluster and all other methods when K_{vert} was greater. Therefore, the sacral marker cluster may not be appropriate for the calculation of K_{vert} during situations where K_{vert} is relatively high, such as on soft surfaces.

The law of falling bodies method demonstrated the least agreement between each pair of methods for the calculation of K_{vert} and ΔCOM_L . Similar to the sacral marker cluster, a proportional systematic error was revealed between the law of falling bodies method when compared to all other methods for the calculation of K_{vert} and ΔCOM_L . Further, the law of falling bodies method had the greatest variance for the calculation of K_{vert} and ΔCOM_L compared to all other methods. Calculation of K_{vert} and ΔCOM_L may not be accurate using the law of falling bodies method due to a number of assumptions which may not be true. The law of falling bodies method assumes maximum vertical velocity of the COM to occur at IC followed by a linear decrease to peak vGRF at which moment there is assumed to be a change in direction and thus velocity to be $0 \text{ m}\cdot\text{s}^{-1}$. However, this has been shown to be inaccurate with a

report that maximum vertical velocity of the COM occurs when vGRF is equal to body mass (Blickhan, 1989) during the loading phase. Therefore, calculated vertical velocity of the COM during loading will be lower, leading to a lower value for ΔCOM_L and higher K_{vert} value.

In contrast, the law of falling bodies and double integration method demonstrated the greatest agreement for the calculation of ΔCOM_f . The near identical results (bias = 0.02 cm) between methods are due to the same underlying principle of acceleration due to gravity being constant during the flight phase. The difference between the two methods being that the double integration method calculates ΔCOM_f on a point-by-point basis while the law of falling bodies method calculated ΔCOM_f as a mean value over the second half of the flight phase.

3.5 Conclusion

The segmental analysis method is considered to represent the gold standard measure of K_{vert} during single-leg, on-the-spot hopping. The calculation of K_{vert} from the segmental analysis method minimises the number of assumptions and has the highest repeatability (lowest variance) across multiple measurements. The double integration method is most comparable to the segmental analysis method and can be used for the accurate estimation of K_{vert} , ΔCOM_L and ΔCOM_f during single-leg hopping. Care should be taken when comparing K_{vert} , ΔCOM_L and ΔCOM_f computed from the sacral marker cluster to other methods. The law of falling bodies method should be avoided for the calculation of K_{vert} and ΔCOM_L without first determining a valid conversion factor.


CHAPTER FOUR


Adaptation of lower limb movement patterns when maintaining performance in the presence of muscle fatigue

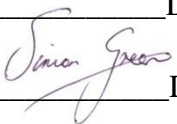
I, Kurt Mudie, hereby declare that I was the principal researcher of all work included in this chapter, including work published with multiple authors. A statement from the authors confirming the authorship and contribution of the PhD candidate to the published work is presented below.

As co-authors of the paper ‘Adaptation of lower limb movement patterns when maintaining performance in the presence of muscle fatigue’, we confirm that Kurt Mudie made the following contributions:

- Conception and design of the research questions
- Data collection, processing and analysis
- Statistical analyses and interpretation of findings
- Writing and review of the chapter/paper
- Corresponding author for communication with journal

Peter J Clothier:  Date: 6/01/17

Amitabh Gupta:  Date: 04/01/17

Simon Green:  Date: 16/01/2017

The following publications have emanated from the study described in Chapter 3:

Mudie, K. L., Gupta, A., Green, S., & Clothier, P. J. (2016). Adaptation of lower limb movement patterns when maintaining performance in the presence of muscle fatigue. *Human Movement Science, 48*, 28-36. doi: 10.1016/j.humov.2016.04.003. IF: 1.598 (Appendix B)

4.1 Introduction

Variability is an inherent characteristic of human movement that occurs at multiple levels of movement organisation (Preatoni et al., 2013). Although low variability in performance output is desirable, variability in the movement between couplings (two joints or segments; Figure 4.1) has been suggested to play a functional role and contribute to a successful performance output during repetitive tasks (Hamill et al., 2012; Latash, 2012; Preatoni et al., 2013). Coupling variability may provide flexibility to the system by permitting adaptation to movement errors or changes in intrinsic or extrinsic factors, such as fatigue or the environment respectively (Bartlett et al., 2007; Hamill et al., 2012; Hamill et al., 1999; Preatoni et al., 2013). Therefore, greater coupling variability is postulated to be beneficial by permitting multiple movement solutions to a specific task (Bartlett et al., 2007; Hamill et al., 2012; Hamill et al., 1999; Preatoni et al., 2013). However, too much or too little coupling variability may be detrimental to the musculoskeletal system and associated with pathology whereby, optimal variability is within the range of these extremes (Bartlett et al., 2007; Hamill et al., 2012; Hamill et al., 1999; Preatoni et al., 2013).

There are conflicting findings in the literature of changes in coupling variability as fatigue progresses during repetitive tasks. Coupling variability of the thigh-shank and the shank-foot did not change during running to fatigue (Miller et al., 2008). In contrast, variability of the hip-knee coupling decreased during a 45° anticipated cut task following an isolated hamstring fatigue protocol (Samaan et al., 2015). Further, variability of the shank-rear foot coupling was shown to increase during treadmill walking following localised fatigue of the tibialis posterior muscle (Ferber & Pohl, 2011). Trunk-thigh and thigh-shank coupling variability also increased during the performance of a repetitive maximal vertical jump test (Dal Pupo et al., 2013).

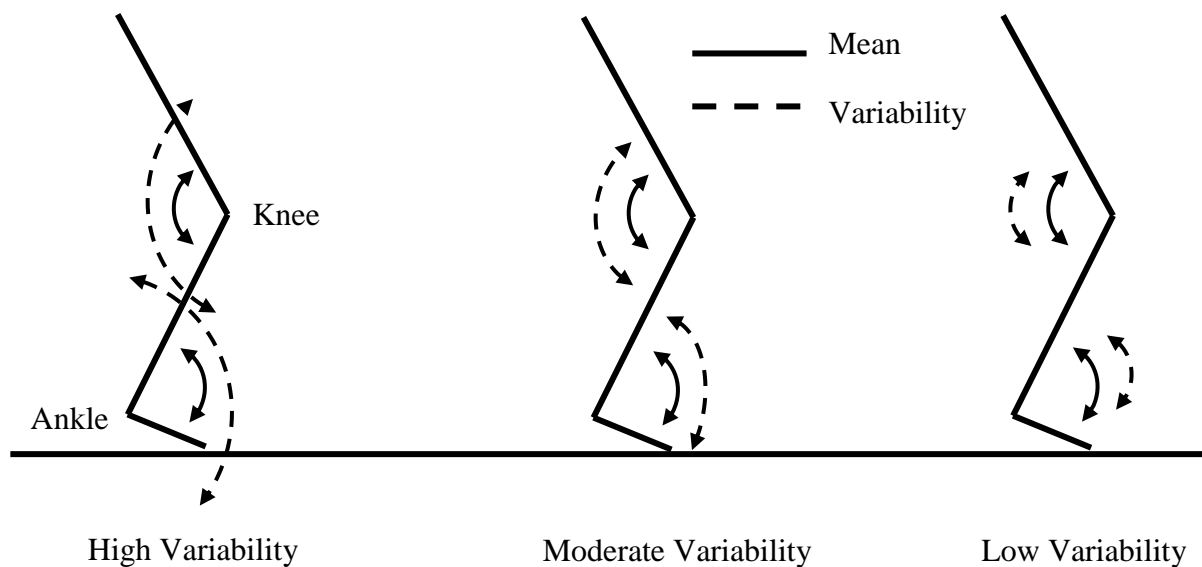


Figure 4.1 Schematic diagram representing mean (solid line) movement for each joint of the knee – ankle coupling over a consecutive number of cycles with the variability (dashed line) overlaid. Although mean joint movement remains the same, coupling variability is high on the left, moderate in the middle and low on the right.

These conflicting findings may be due to the measure of variability being sensitive to either, differences in tasks between studies or changes in the performance output that would likely occur during a fatiguing task (Miller et al., 2008). Thus, performance output characteristics that modulate leg stiffness and were not reported in previous studies, such as stride length (Miller et al., 2008), force output (Samaan et al., 2015) or jump height (Dal Pupo et al., 2013), may have changed as fatigue increased and may have directly affected the measurement of coupling variability. Further, it is difficult to differentiate possible effects due to warm-up, motor learning during repetitive and novel tasks or fatigue, when coupling variability was not measured regularly (Dal Pupo et al., 2013; Ferber & Pohl, 2011; Miller et al., 2008; Samaan et al., 2015) during a repetitive task.

It remains unclear as to whether the reported changes in coupling variability were due to changes in task performance or in fact due to fatigue. The purpose of this study was to examine the effect of local muscle fatigue on coupling variability when performance output was maintained during a repetitive loading task. It was

hypothesised that coupling variability would increase when performance output was maintained in the presence of increasing muscle fatigue.

4.2 Methods

4.2.1 Participants

Forty recreationally active men ($n = 20$) and women ($n = 20$) (mean \pm standard deviation (SD) 22.7 ± 3.0 years of age; 1.7 ± 0.1 m in height; 68.8 ± 10.7 kg in mass) volunteered to participate in this study. All participants were healthy and reported participating in exercise for between 1 to 4 hours per week. Participants reported no past or current history of lower limb pathology, injury, pain or lower limb fracture within the six months prior to testing. Participants were excluded if they had a history of lower limb surgery(s). Ethical approval (H1074) was granted by the University of Western Sydney Human Research Ethics Committee (Appendix F). Prior to testing, all participants read through information describing the study (Appendix G), completed a pre-exercise health screen questionnaire (Appendix H) and provided written and informed consent (Appendix I).

4.2.2 Instrumentation

Kinetic and kinematic data were collected synchronously during single-leg, on-the-spot hopping to volitional exhaustion. The instrumentation used to collect the kinetic and kinematic data were described in Chapter 3 (Section 3.2.2). All data were collected and time synchronised using First Principles (Version 1.2.4), and later processed using Visual 3D (C-Motion, Version 4, Germantown, MD).

4.2.3 Participant preparation

Participants wore high-cut running shorts and men were shirtless whilst women wore a tight fitting crop top. Following measurement of each participant's height and body mass, a warm-up and a hopping familiarisation period was completed (Hobara, Kobayashi, et al., 2013).

To control task performance of K_{vert} , a target hopping frequency and height were imposed. Consistent with previous research (Farley et al., 1991; Hobara, Kobayashi, et al., 2013) each participant was instructed to land on the audible tone of a

metronome set to 2.2 Hz (132 beats/minute). Tactile feedback was provided to allow each participant to hop to a predetermined target height by a custom-built apparatus on which 32 mm wide elastic bands were stretched horizontally above the participant's head (Section 4.2.4). All trials were performed barefoot (Hobara et al., 2011; Zuur et al., 2010) and on the participant's dominant leg, defined as their preferred kicking leg (Hobara, Inoue, et al., 2013; Padua et al., 2006). Each participant was instructed to keep their hands on their hips, land on the audible tone of the metronome on the ball of their foot and lightly touch their head on the elastic bands when hopping to the target height.

4.2.4 Controlling task performance

An audible cue and tactile feedback were provided with the aim to control the cadence of hopping and vertical displacement of the COM, respectively. The use of an audible metronome has been shown to lead to maintenance of hopping cadence (Farley et al., 1991; Hobara et al., 2011). However, the determination of K_{vert} has been performed using a number of different methods. In the current study it was imperative to maintain K_{vert} and therefore a preferred measure of vertical displacement of the COM and K_{vert} was determined (Mudie, Gupta, Green, Hobara, & Clothier, 2016) (Chapter 3).

Briefly, four methods including two methods using kinetic data and two methods using both kinematic and kinetic data were evaluated and compared as estimates of measuring vertical displacement of the COM and K_{vert} during on-the-spot single-leg hopping. Of the four methods evaluated, the segmental analysis method, which used the coordinates and segment masses of the seven segments, was determined to have the least assumptions and demonstrated the highest repeatability for the calculation of vertical displacement of the COM and K_{vert} . Consequently, the segmental analysis method was used as the method to calculate vertical displacement of the COM and K_{vert} as a descriptor of performance output.

To control task performance, target hop height was calculated for each participant by recording a trial of single-leg hopping at a cadence of 2.2 Hz for 15 s. Vertical displacement of the COM during the flight phase was calculated for each hop cycle using the segmental analysis method (see sections 4.2.6 – 4.2.7 for detailed description of the processing and analysis). Target hop height was set to each

participant's preferred hopping height at 2.2 Hz, calculated as the mean vertical displacement of the COM during the flight phase from the sixth to the tenth hop cycle (Hobara, Inoue, et al., 2013; Hobara, Kobayashi, et al., 2013). The elastic bands were then set at the target hop height using a laser measurement device (Bosch PLR 50, measurement accuracy ± 2 mm).

4.2.5 Testing protocol

A static calibration trial with the participant standing in the anatomical position was recorded. Each participant then completed a single effort of on-the-spot single-leg hopping to volitional exhaustion at 2.2 Hz to a target hop height. Volitional exhaustion was defined as the moment when the participant could no longer maintain the required performance output of hopping frequency or target hop height (Morio et al., 2011; Nicol et al., 2006; Regueme et al., 2005). To quantify the acute functional effects at volitional exhaustion, five continuous maximal height single-leg hops were performed three minutes before (PRE) and 10 s after (POST) the exhaustive single-leg hopping trial.

4.2.6 Data processing

All recorded trials were exported from First Principles software (Version 1.2.4) as C3D files for processing (Visual 3D, Version 4). Kinematic marker recordings were interpolated using spline interpolation for up to a maximum gap of 10 frames and filtered using a fourth order zero-lag bidirectional low-pass Butterworth filter with an 8 Hz cut-off (Bobbert & Richard Casius, 2011; Hobara et al., 2011; Hobara et al., 2008). Force plate data were filtered using a fourth order zero-lag bidirectional low-pass Butterworth filter with a 50 Hz cut-off frequency (Gupta et al., 2016; Gupta et al., 2014), respectively.

4.2.7 Data analysis

A single hop cycle was defined as a complete flight phase and the subsequent contact phase (Figure 3.6). The contact phase included the loading (initial contact to peak vGRF) and propulsive (peak vGRF to toe-off) phase. Temporal events including initial contact and toe off were determined for each hop cycle from the vGRF trace as the first and last data point ≥ 10 N (Lloyd et al., 2009), respectively. Peak vGRF was

defined as the maximal vGRF value during the contact phase. Contact time represented the time the foot was in contact with the force plate and was calculated as the duration between initial contact and toe-off for each hop cycle (Lloyd et al., 2009).

Dependent variables calculated for each hop cycle during the exhaustive trial included hopping frequency, hopping height, contact time, vertical stiffness and coupling variability of the knee-ankle and hip-knee couplings. Dependent variables were calculated as the mean of 10 consecutive hop cycles (Kuitunen, Ogiso, & Komi, 2011; Maton & Pellec, 2001; Padua et al., 2006) at time periods of 0, 20, 40, 60, 80 and 100% of the duration of the trial (Microsoft Office Excel, 2007 and MATLAB, 2012B 32-bit). To quantify the acute functional effects at volitional exhaustion, the single-hop cycle with the greatest hop height was chosen during the PRE and POST maximal effort hop trials and compared.

4.2.7.1 Performance output

Maximal vertical displacement of the COM during the flight (ΔCOM_f) and loading (ΔCOM_L) phases were determined by a segmental analysis method (Mudie et al., 2016; Ranavolo et al., 2008) (Chapter 3). Vertical stiffness (K_{vert}) was calculated as the quotient of force normalised to body mass and COM displacement during the loading phase (Equation 4.1) (Butler et al., 2003; Farley et al., 1998; Farley & Morgenroth, 1999; Ferris & Farley, 1997; Serpell, Ball, Scarvell, & Smith, 2012).

$$K_{\text{vert}} = \frac{\text{peak vGRF}}{\Delta\text{COM}_L} \quad (4.1)$$

where K_{vert} was vertical stiffness ($\text{N}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$), peak vGRF was the maximum vGRF during the contact phase normalised to body mass ($\text{N}\cdot\text{kg}^{-1}$) and ΔCOM_L was the loading height (m).

4.2.7.2 *Joint coupling variability*

Intra-limb joint coupling variability is a linear tool that investigates the magnitude of movement variability (Hamill et al., 2012; Preatoni et al., 2013), calculated to describe the variability in coordination between the knee and ankle joints and hip and knee joints. Joint coupling variability was calculated for knee flexion/extension - ankle flexion/extension (KxAx), hip flexion/extension - knee flexion/extension (HxKx) and knee flexion/extension - ankle eversion/inversion (KxAy) during contact with the force plate. These couplings were chosen due to the predominant unidirectional vertical motion of hopping in the sagittal plane, the role of the ankle and knee joints for modulating submaximal hopping (Lamontagne & Kennedy, 2013) and the kinetic chain between lower limb joints. Quantification of coupling variability was calculated using a modified vector coding technique (Ferber, Davis, & Williams III, 2005; Heiderscheit et al., 2002; Pollard et al., 2005). Initially, all joint angle data during the hop cycle loading and propulsive phases were normalised to 101 data points (Ferber et al., 2005; Heiderscheit et al., 2002; Pollard et al., 2005). Motion of the proximal and distal joints were then plotted on the ordinate and abscissa, respectively (Ferber et al., 2005). The coupling angle (Θ) was calculated as the orientation of the resultant vector to the right horizontal between two adjacent data points as follows (Figure 4.2; Equation 4.2):

$$\Theta_i = \text{abs} [\tan^{-1} (y_{i+1} - y_i / x_{i+1} - x_i)] \quad (4.2)$$

where i = data point 1, 2 and n of the time series (Ferber et al., 2005; Heiderscheit et al., 2002; Pollard et al., 2005). Calculated values were then converted from radians to degrees with the resultant coupling angles ranging from 0 – 90° (Ferber et al., 2005; Pollard et al., 2005). A coupling angle of 45° indicated equal motion from both proximal and distal joints. A coupling angle greater or less than 45° indicated greater motion in the proximal or distal joint than the other joint, respectively (Ferber et al., 2005).

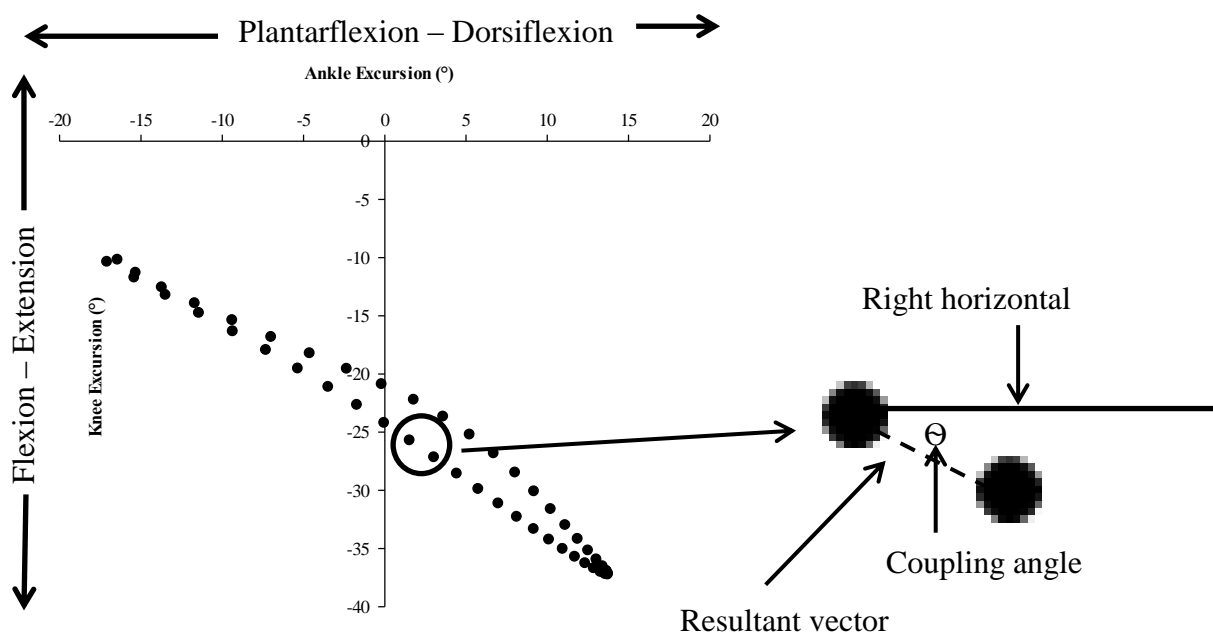


Figure 4.2 Vector coding technique used to calculate coupling angles (Θ) between two joints. Plotted knee and ankle excursions during the contact phase (left) and an enlarged image of two subsequent data points with Θ indicated (right).

To calculate the mean joint coupling angle across the loading and propulsion phases, the mean coupling angle between each data point was calculated on a point-by-point basis across 10 consecutive hop cycles, resulting in a mean for each of the 100 time points over each phase (Figure 4.3) (Ferber et al., 2005; Heiderscheidt et al., 2002; Pollard et al., 2005). At each data point, the SD of the coupling angle across the 10 hop cycles was calculated, resulting in a 100 point variability time series for the loading and propulsion phases (Figure 4.4) (Ferber et al., 2005; Heiderscheidt et al., 2002; Pollard et al., 2005). The mean joint coupling variability during the loading and propulsion phases was calculated as the mean value of the variability time series during each phase, providing a measure of between-cycle, within-participant variability (Figure 4.4) (Ferber et al., 2005; Heiderscheidt et al., 2002; Pollard et al., 2005). Joint coupling variability from 10 consecutive hop cycles was determined for the loading and propulsive phases at 0, 20, 40, 60, 80 and 100% of the trial (Ferber et al., 2005; Heiderscheidt et al., 2002; Pollard et al., 2005).

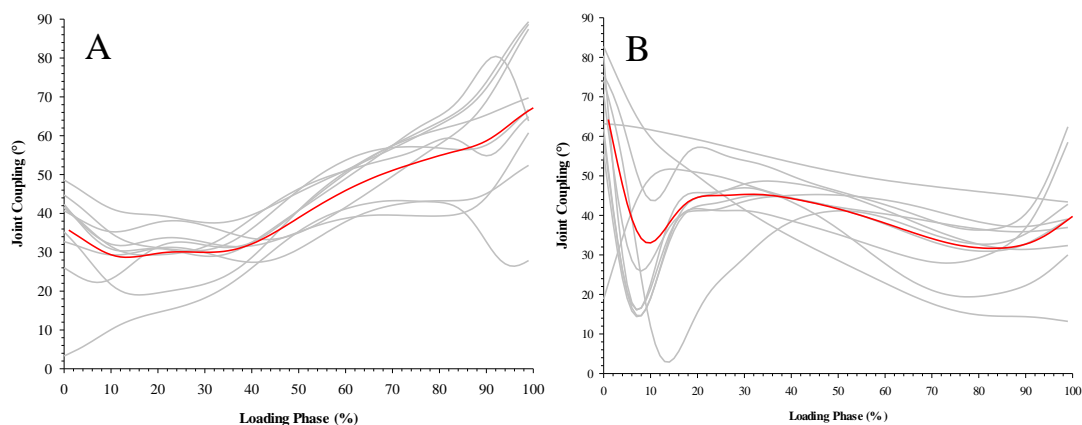


Figure 4.3 Knee-Ankle joint coupling ($^{\circ}$) as a function of the (A) loading and (B) propulsion phase (0 – 100%) for a single representative male participant. The 10 dashed grey lines represent the calculated joint coupling angle for the first 10 hop cycles of the trial. The single red line represents the calculated mean joint coupling angle of the first 10 hop cycles as a function of the (A) loading and (B) propulsion phase (0 – 100%).

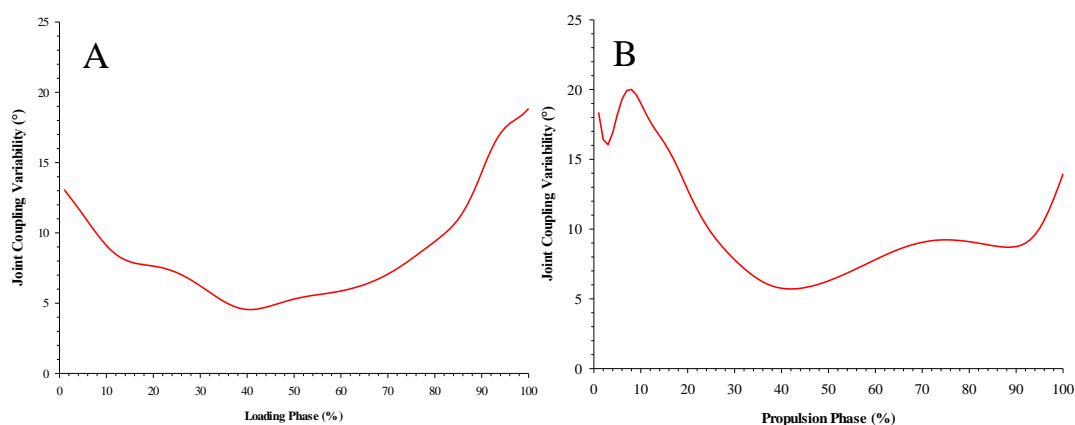


Figure 4.4 Knee-Ankle joint coupling variability ($^{\circ}$) as a function of the (A) loading and (B) propulsion phase (0 – 100%) for a single representative male participant. Each time point (0 – 100%) was calculated as the standard deviation of the coupling angle at that time point across the first 10 hop cycles, resulting in a 100 point variability time series across the (A) loading and (B) propulsion phase, providing a measure of between-cycle, within-participant variability.

4.2.8 Statistical analyses

A one-way (time period) ANOVA with repeated measures (0, 20, 40, 60, 80 and 100% time periods) was performed to compare all dependent variables during the exhaustive trial (SPSS, Version 22). Mauchly's test of sphericity was performed and if violated ($p < 0.05$) a Greenhouse-Geisser correction procedure was used. *Post-hoc* pairwise multiple comparisons were made between the 0% time period and subsequent time periods with a Bonferroni correction applied. To compare maximal hop height between PRE and POST trials a two-tailed paired sample *t*-test was performed. The Alpha level was set *a priori* at $p < 0.05$ for all statistical analyses.

4.3 Results

4.3.1 Fatigue and performance output

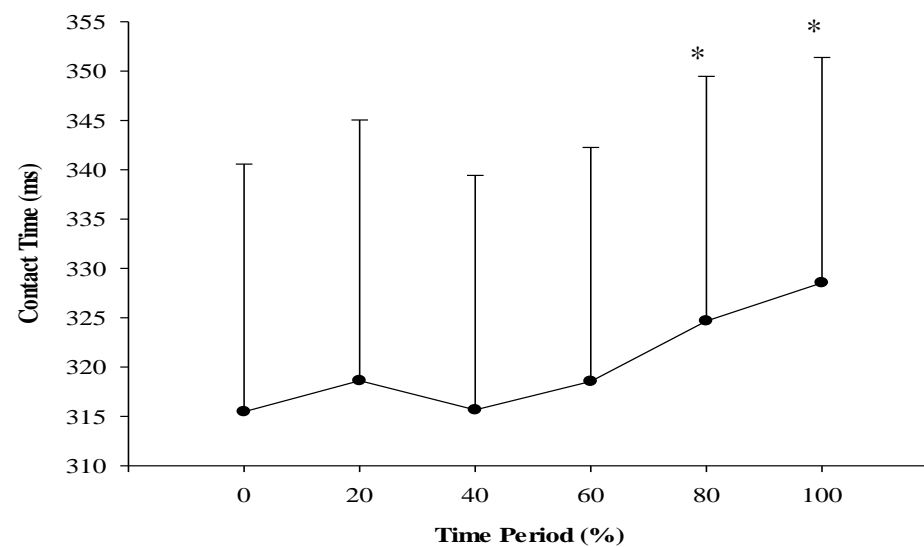
Total duration of single-leg hopping (mean \pm SD) to volitional exhaustion was 79 ± 27 s with a target hop height of 2.50 ± 1.08 cm (range 0.65 to 5.00 cm). There was a significant decrement ($p < 0.001$) of $22 \pm 16\%$ in maximum hopping height (PRE: 9.38 ± 2.81 cm; POST: 7.23 ± 2.49 cm) following the exhaustive hopping trial (Appendix L.1). Further, there was a significant increase in contact time during the exhaustive hopping trial from the 0% time period to the 80% ($p = 0.044$) and 100% ($p = 0.007$) time periods (Figure 4.5). In contrast, there was no significant change detected for performance output characteristics, including hopping frequency, hopping height and vertical stiffness between each time period (Appendix L.2; Table 4.1).

4.3.1 Joint coupling variability

During the loading phase, there was a significant increase in K_{xAx} and K_{xAy} between the 0% time period and the 80% ($p = 0.002$; $p = 0.005$) and 100% ($p < 0.001$; $p < 0.001$) time periods. There was a significant increase in H_{xKx} during the loading phase between the 0% time period and the 40% ($p = 0.011$), 60% ($p < 0.001$), 80% ($p < 0.001$) and 100% ($p < 0.001$) time periods (Figure 4.6; Appendix L.3). During the propulsion phase, there was a significant increase in K_{xAx} between the 0% time period and the 100% ($p < 0.001$) time period, and H_{xKx} between the 0% time period and the 80% ($p = 0.002$) and 100% ($p < 0.001$) time periods. In contrast, no significant difference was detected for K_{xAy} during propulsion between the 0% time period and any subsequent time period (Figure 4.6; Appendix L.4).

Table 4.1 Performance output characteristics (mean \pm SD) at all time periods during the exhaustive single-leg hopping trial.

	0%	20%	40%	60%	80%	100%
Hopping Frequency (Hz)	2.23 \pm 0.05	2.21 \pm 0.03	2.22 \pm 0.04	2.23 \pm 0.04	2.21 \pm 0.05	2.23 \pm 0.05
Hopping Height (cm)	1.82 \pm 0.85	1.83 \pm 0.96	1.90 \pm 0.84	1.82 \pm 0.84	1.80 \pm 0.88	1.71 \pm 0.77
Vertical Stiffness (N \cdot kg ⁻¹ \cdot m ⁻¹)	247.88 \pm 23.48	243.11 \pm 21.88	245.41 \pm 20.20	244.20 \pm 23.16	238.51 \pm 25.14	237.22 \pm 21.70

**Figure 4.5** Mean (+SD) contact time (ms) at each time period over the duration of hopping. * indicates $p < 0.05$ compared to the 0% time period.

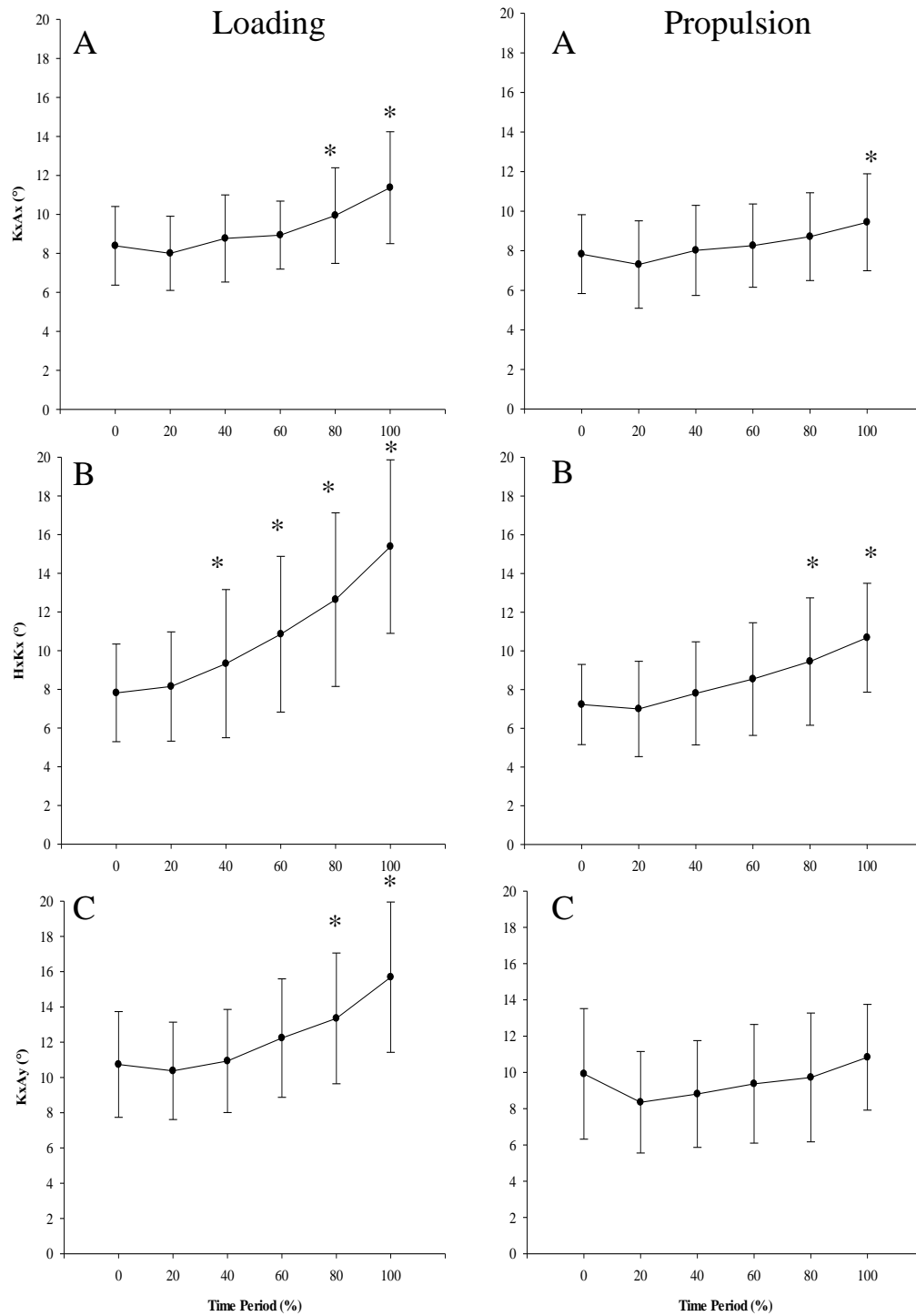


Figure 4.6 Mean (\pm SD) coupling variability ($^{\circ}$) during loading (left) and propulsion (right) at each time period for: A) knee flexion/extension – ankle flexion/extension (KxAx); B) hip flexion/extension – knee flexion/extension (HxKx); and C) knee flexion/extension – ankle abduction/adduction (KxAy). * indicates $p < 0.05$ compared to the 0% time period.

4.4 Discussion

This study examined adaptations in joint coupling variability when performance output was maintained as local muscle fatigue increased during a repetitive loading task. The results of the current study supported the alternate hypothesis, that there was an increase in joint coupling variability when performance output was maintained with increasing muscle fatigue. This suggests that during a fatiguing rapid and repetitive loading task, healthy individuals were able to maintain a required performance level with multiple movement patterns.

The single-leg hopping protocol to exhaustion was considered to be an adequate stimulus to elicit an increase in muscle fatigue of the lower limb muscles. This was supported by the finding of a significant reduction in post-trial maximum hopping height and an increase in contact time during the trial performance, consistent with previous research on the effect of fatigue on a repeated jump task (Dal Pupo et al., 2013; McNeal, Sands, & Stone, 2010; Morio et al., 2011; Morio et al., 2012).

A key difference between previous studies and the current study was that the task performance was maintained such that vertical stiffness remained unchanged throughout the entire trial by imposing a target hopping cadence and height (Farley et al., 1991; Ferris & Farley, 1997; Hobara et al., 2011). Performance was maintained despite the onset of and increasing fatigue during the trial. The capacity to withstand increasing fatigue while maintaining performance could be explained by the notion that limb multi-segment coordination may alter even though the task requirements remained unchanged. Inherent to this is evidence that the local muscle fatigue of the triceps surae does lead to a reduced ability to generate maximal power output (Dal Pupo et al., 2013). Due to the multi-segment coordinative strategies available, it is plausible that there is an inability to generate adequate ankle plantarflexion moment during contact, such that there were changes in the contribution of the thigh and hip muscles that produced compensatory movements. It is speculated that these compensatory movements of the thigh and hip muscles accommodated the required ankle joint power when the ankle plantarflexors were working both eccentrically during the loading phase and concentrically during the propulsion phase to maintain performance.

Lower limb compensatory movements are reflective of changes in variability in the movement between two joints or segment couplings. Findings in the current study demonstrate significant increases in HxKx and KxAx coupling variability towards the end of the trial when fatigue was more evident, indicated by a greater ground contact time. The current results support previous findings of increased hip-knee and knee-ankle coupling variability when fatigued (Dal Pupo et al., 2013; Ferber & Pohl, 2011).

In the current study there were increases in coupling variability during both the loading and propulsive phases in hopping. In contrast to previous results during a repetitive maximal vertical jump test (Dal Pupo et al., 2013), the current study demonstrated that during hopping increases in coupling variability during the loading phase occurred earlier and were of higher magnitude than during the propulsive phase (Figure 4.6). The contrasting results may be due different movements tasks completed and that the current task was submaximal, compared to the maximal effort task by Dal Pupo et al. (2013). As fatigue progresses during a maximal effort task, changes are suggested to occur more in characteristics that directly impact performance, such as during propulsion to achieve a maximal jump height (Nicol et al., 2006). Whilst during a submaximal task, compensatory strategies that allow performance to be maintained are suggested to be more evident, such as during the loading phase (Nicol et al., 2006). Thus, it is plausible that the observed increase in variability during the current submaximal task was a compensatory strategy that may be used to withstand the effects of fatigue enabling the individual to maintain the required performance output. Maintaining performance despite an increase in coupling variability during the loading phase may also support the functional role of coupling variability during repetitive loading tasks.

The observed increase in coupling variability during loading in this study, may be explained by the greater tension of lower limb musculature that assists to decelerate the body in preparation for the upward propulsive phase (Cavagna, 1977). In contrast to the loading phase, the propulsive phase may display lower levels of muscle activity with a greater reliance on utilising stored elastic energy (Lamontagne & Kennedy, 2013). Thus, greater stress on lower limb musculature can be experienced attenuating the landing forces compared to the forces generated during the upward propulsion. Increased coupling variability during the loading phase may enable the

landing forces to be more optimally attenuated by the individual distributing the load across various structures and/or muscles (Hamill et al., 1999; Miller et al., 2008).

This increased coupling variability may permit flexibility to the lower limb to adapt to increases in local muscle fatigue, which may affect the landing (Pollard et al., 2005; Wilson et al., 2008).

In addition to the lower increase in coupling variability during the propulsive phase, the increase in the variability of the knee-ankle couplings was also lower in magnitude than the increase in hip-knee coupling variability. Specific to the current task, joint motions of predominantly the ankle and knee during the propulsive phase were characteristics that were observed to modulate submaximal hopping height (Dal Pupo et al., 2013; Lamontagne & Kennedy, 2013). With increasing fatigue of the lower limb muscles, the increase of knee-ankle coupling variability was not associated with any change in performance. Although coupling variability is beneficial during repetitive tasks, too much variability in characteristics that modulate performance may cause the motor system to be unstable and affect the ability to achieve the required performance (Hamill et al., 2012). Therefore, increases in coupling variability may be limited by the constraints of the task and whether the task is able to be maintained at a constant output. Thus, as the task of hopping was predominantly modulated by the knee-ankle coupling, a relatively large magnitude of variability in the hip-knee coupling might allow enough source of variability in the kinetic chain and still maintain a consistent performance.

As the task progressed variability of the HxKx coupling increased, with a significant increase occurring early in the trial from the 40% time period onwards. A subsequent and significant increase in the KxAx coupling occurred onwards from the 80% period. In a task where there is a required performance output goal, a change in control of one element will likely require a reactive compensation from another in an attempt to maintain the required goal of the movement (Latash et al., 2010). Thus, the latter significant increase in variability of the KxAx coupling may be a reactive compensatory mechanism to both increased muscle fatigue whilst maintaining the performance output of the task. The current findings support previous research, highlighting that compensatory mechanisms are observed during other fatiguing and repetitive tasks when maintaining performance such as finger tapping (Singh, Varadhan, Zatsiorsky, & Latash, 2010), postural sway (Singh & Latash, 2011) and

walking (Ferber & Pohl, 2011). The results of the current study reveal that appropriate compensatory strategies are also present during a rapid and repetitive loading task in the lower limb.

4.5 Conclusion

Increased coupling variability may allow greater available movement solutions between the coupled joints as muscle fatigue progresses. This may be an effective strategy to overcome the effects of increased local muscle fatigue by permitting alternate movement patterns with the maintenance of motor performance. Alternate movement patterns which would distribute the load across various muscles may delay the detrimental effects of fatigue and preserve performance output. This finding supports the notion that performance is able to be preserved despite the presence of variability in the neuromotor system.


CHAPTER FIVE


Adaptations in neuromuscular activity during a repetitive loading task to exhaustion

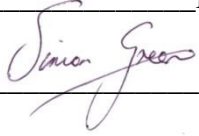
I, Kurt Mudie, hereby declare that I was the principal researcher of all work included in this chapter. A statement from the co-authors confirming contribution of the PhD candidate to the work is presented below.

As co-authors of the chapter ‘Adaptations in neuromuscular activity during a repetitive loading task to exhaustion’, we confirm that Kurt Mudie made the following contributions:

- Conception and design of the research question
- Data collection, processing and analysis
- Statistical analyses and interpretation of findings
- Writing and review of the chapter

Peter J Clothier:  Date: 6/01/17

Amitabh Gupta:  Date: 04/01/17

Simon Green:  Date: 16/01/2017

5.1 Introduction

To maintain a consistent performance output during a repetitive task, there may be multiple compensatory mechanisms employed by an individual. The finding that joint coupling variability changed as on-the-spot, single-leg hopping progressed to exhaustion, demonstrated that movement variability was fluctuant even while leg stiffness remained unchanged (Chapter 4). It may be inferred that these joint kinematic adjustments occurred due to changes in the timing and relative activity of the lower limb muscles, which control multi-segment lower limb motion. There have been demonstrated changes in both the feedforward (central) and feedback (spinal-reflex) activity of the triceps surae when there are adjustments in vertical stiffness (Farley et al., 1998; Farley & Morgenroth, 1999; Hobara et al., 2007; Hortobágyi & DeVita, 2000). However, it is unclear how muscle activity changes during an exhaustive task when there are no global changes to leg mechanical characteristics.

Different compensatory strategies following a fatiguing task involving the lower limb have been previously observed. For example, significant decreases in MG and SOL muscle activity following 60 s of maximal height hopping (Moritani et al., 1990) and submaximal height rebounds to exhaustion on a sledge apparatus (Kuitunen et al., 2007) have been demonstrated. In contrast, significant increase in MG and SOL muscle activity was demonstrated during double-leg hopping, following a fatiguing squat exercise (Padua et al., 2006). These studies highlight the variability that may be present following different testing and fatiguing protocols.

Between subject differences following the same protocol may also occur as demonstrated in four volleyball players who performed double-leg hopping to exhaustion at 120 hops per minute (Bonnard et al., 1994). Although only four participants were included in the study by Bonnard et al. (1994), two participants were identified to have an earlier activation of the MG and no change in the iEMG of the MG. In contrast, the authors observed that the other two participants had a decrease in the iEMG of the MG in the feedforward period and in the eccentric phase of the hopping cycle. The differing responses between participants was suggested to be part of a change in muscle synergies with an increase in VL activity in participants with the decrease in MG activity, which was not observed in the other

two participants (Bonnard et al., 1994). The lack of statistical power of only four participants makes it difficult to ascertain any conclusions.

The conflicting findings within and between studies may be due to both time-dependent (Morio et al., 2011; Regueme et al., 2005) and task-dependent influences (Komi, 2000; Nicol et al., 2006) affecting neuromuscular activity. For example, in studies which recorded neuromuscular activity during hopping there were changes in hop height (Kuitunen et al., 2007; Moritani et al., 1990; Padua et al., 2006) and vertical stiffness (Kuitunen et al., 2007; Moritani et al., 1990). Both feedforward and feedback responses are implicated to be affected by any change in task performance (Santello, 2005; Santello & McDonagh, 1998). Therefore, it is difficult to determine whether the changes in neuromuscular activity during a repetitive task to exhaustion are due to fatigue and/or a change in performance output. Considering non-linear changes in EMG temporal and amplitude characteristics have been observed over the time course of an exhaustive sledge rebound task (Morio et al., 2011; Regueme et al., 2005) it is important to evaluate the changes in neuromuscular activity over the course of a repetitive loading task to exhaustion.

Adaptations in neuromuscular activity remain unclear during an exhaustive repetitive loading task when performance output is maintained throughout the entire trial. The purpose of this study was to evaluate the effect of fatigue on changes in neuromuscular activity when performance output was maintained during on-the-spot single-leg hopping. It was hypothesised there would be a latency in temporal characteristics and an increase in amplitude characteristics of lower limb muscle activity during the performance of on-the-spot single-leg hopping to volitional exhaustion.

5.2 Methods

Data which was analysed to test the hypothesis in this study had been collected from the same participants reported in Chapter 4. Methods common to the current Chapter and Chapters 3 & 4, which have been previously described in detail, are presented here in brief.

5.2.1 Participants

Forty recreationally active men ($n = 20$) and women ($n = 20$) (mean \pm SD 22.7 ± 3.0 years of age, 1.7 ± 0.1 m in height and 68.8 ± 10.7 kg body mass) volunteered to participate in this study. All participants were healthy and reported no past or current history of lower limb pathology within the six months prior to testing. Ethical approval (H1074) was granted by the University of Western Sydney Human Research Ethics Committee (Appendix F) and all participants provided written and informed consent prior to testing (Appendix I).

5.2.2 Instrumentation

Kinetic, kinematic and EMG data were collected synchronously during single-leg, on-the-spot hopping to volitional exhaustion. The instrumentation used to collect the kinetic and kinematic data were described in Chapter 3 (Section 3.2.2).

Electromyography data were collected from seven wireless surface electromyography (sEMG) units (1500 Hz, Aurion, Zerowire, Italy). All data were time synchronised, recorded (First Principles, Version 1.2.4) and post-processed offline following data collection (Visual 3D, C-Motion, Version 4, Germantown, MD).

Surface electromyography signals were transmitted from seven wireless probes to a base unit (Aurion, Zerowire, Italy) at a frequency ranging between 2400 – 2524 MHz. The base unit amplified the signals with an input impedance of 20 M Ω , common mode rejection ratio of 90 dB, signal-noise ratio of > 50 dB and a gain of 1000. The sEMG analogue signals were then transmitted from the base unit to a multiplexer board by 16 BNC connectors. A 50 pin ribbon cable (National Instruments, Australia) transmitted the analogue signals from the multiplexer board to an Optotrak Data Acquisition Unit II (ODAU II), which plugged into the Optotrak System Control Unit, allowing temporal synchronisation of all kinetic, kinematic and sEMG data.

5.2.3 Participant preparation

Participants wore high-cut running shorts and men were shirtless whilst women wore a tight fitting crop top. Following measurement of each participant's height and body mass, a warm-up and familiarisation period was completed barefoot.

The site for placement of sEMG electrode pairs (Maxensor, Medimax Global, Australia) was identified by palpation of the muscle during an isometric contraction and was consistent with SENIAM guidelines for placement of sEMG electrodes (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000) (Appendix M). Surface EMG electrode pairs were adhered over each of seven lower limb muscles (gluteus medius (GM), vastus lateralis (VL), biceps femoris (BF), medial gastrocnemius (MG), soleus (SOL), tibialis anterior (TA) and fibularis longus (FL)) of the hopping leg (Hermens et al., 2000; Hobara et al., 2007; Winter & Yack, 1987) (Appendix M). The wireless sEMG units were button clipped to the electrodes and hypa-fix tape applied to secure the probes and electrodes to the skin during the trial. Crosstalk between adjacent muscles was checked with movement tests of the joint over which the muscle acted (Appendix M). The primary investigator visually inspected the sEMG signal (First Principles, Version 1.2.4) and aimed to identify any crosstalk between muscles. If there was cross-talk then the site for sEMG electrode was adjusted to minimise cross-talk.

5.2.4 Controlling task performance

For the hopping task, an audible metronome was fixed at 2.2 Hz (132 beats per minute) (Farley et al., 1991; Hobara, Kobayashi, et al., 2013) and the target hop height was fixed for each participant as their preferred hopping height at 2.2 Hz. Tactile feedback was provided by a custom build apparatus to allow each person to hop to their target hopping height.

5.2.5 Testing protocol

A static calibration trial was recorded then each participant completed a single effort of on the spot, single-leg hopping to volitional exhaustion at 2.2 Hz to a target hop height. Five continuous maximal effort single-leg hops were performed three minutes before (PRE) and 10 s after (POST) the exhaustive single-leg hopping trial.

5.2.6 Data processing

All recorded data were exported (First Principles, Version 1.2.4) as C3D files for processing (Visual 3D, Version 4). Force plate data were filtered using a fourth order bidirectional low-pass Butterworth filter with a 50 Hz cut-off (Gupta et al., 2016; Gupta et al., 2014). A 13 ms sEMG electromechanical delay was corrected for and the sEMG signals were filtered with a fourth order, bidirectional band-pass filter (50 to 500 Hz) and full wave rectified (Gupta et al., 2014).

5.2.7 Data analysis

A single hop cycle was defined as a complete flight phase and subsequent contact phase determined from the vGRF trace (Figure 3.6). Dependent variables calculated for each hop cycle during the exhaustive trial included hopping frequency (Hz), ΔCOM_f (cm), contact time (s), K_{vert} ($\text{N}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$) and temporal and amplitude muscle activation characteristics derived from sEMG signals. Dependent variables were calculated as the mean of 10 consecutive hop cycles (Kuitunen et al., 2011; Maton & Pellec, 2001; Padua et al., 2006) at time periods of 0, 20, 40, 60, 80 and 100% of the duration of the trial (Microsoft Office Excel, 2007 and MATLAB, 2012B 32-bit). To quantify the acute functional effects at volitional exhaustion, the single-hop cycle with the greatest hop height was chosen during the PRE and POST maximal effort hop trials.

5.2.7.1 *Performance output*

Vertical stiffness, ΔCOM_f and ΔCOM_L were determined by a segmental analysis method (Chapter 3) (Mudie et al., 2016).

5.2.7.2 *Neuromuscular characteristics*

5.2.7.2.1 Temporal characteristics

All temporal characteristics of each muscle were determined by detecting the onset and offset of activity (Figure 5.1). Onset and offset of muscle activity was defined as the instant at which amplitude was at least 10 % of the peak activation amplitude as a mean value over a 50 ms epoch (Gupta et al., 2014) (Figure 5.1). The peak amplitude was defined as the maximum mean amplitude value (MAV) determined

over a moving 50 ms epoch during the loading phase of the hop cycle (Albertus-Kajee, Tucker, Derman, Lamberts, & Lambert, 2011). Subsequently, the minimum value within each 10% epoch before and after the peak activity was labelled as the onset and offset for each lower limb muscle, respectively (Gupta et al., 2014).

Dependent variables calculated for each hop cycle were the duration of the feedforward period (from onset of muscle activity to initial contact) and total duration of muscle activity (from onset to offset of muscle activity) (Figure 5.1).

There was no consistently identified onset or offset for the TA or BF muscles during individual hop cycles (Figure 5.2). Therefore, the duration of the feedforward phase was only calculated for the GM, VL, MG, SOL and FL muscles and total duration for the GM, VL, MG, SOL, FL and TA muscles.

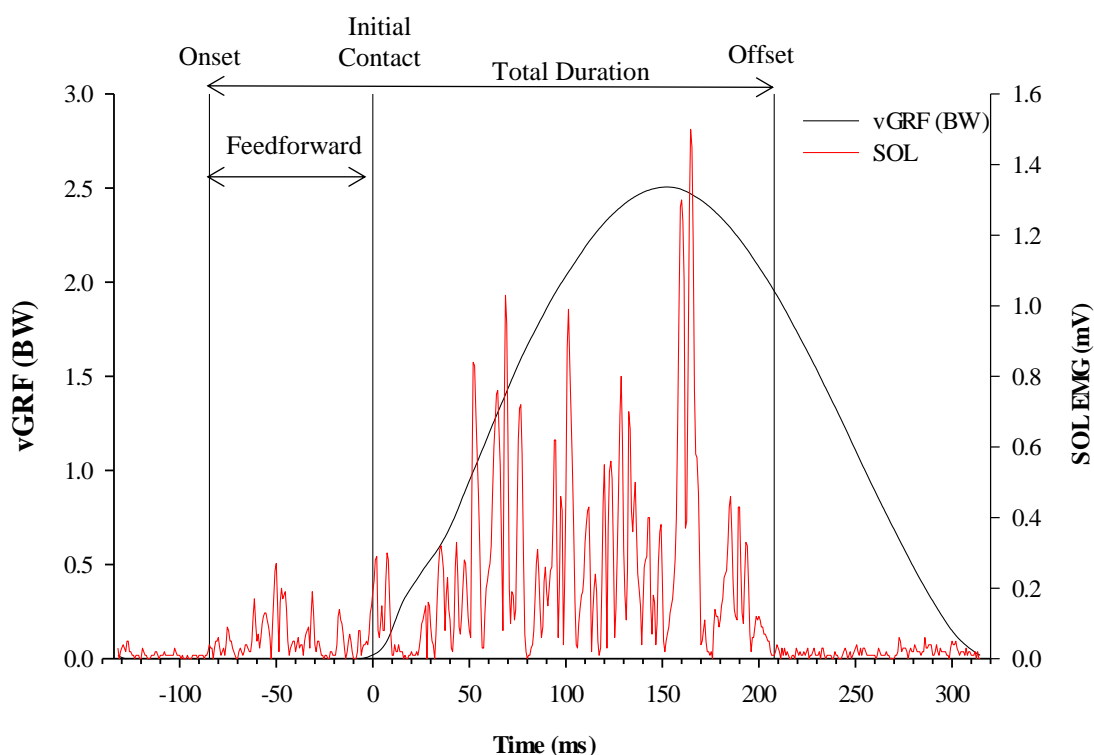


Figure 5.1 Temporal characteristics (duration of feedforward activity and muscle activity from onset to offset) for muscle activity during one hop cycle with a representative soleus (SOL) and vertical ground reaction force (vGRF) trace

5.2.7.2.2 Amplitude characteristics

To calculate muscle activation amplitude characteristics, the MAV for the GM, BF, VL, MG, SOL, FL and TA muscles were calculated over six predefined phases during each hop cycle (Figure 5.2). The six predefined phases encompassed activity before and after initial contact to evaluate both feedforward (central) and feedback (spinal-reflex) activity (Dyhre-Poulsen, Simonsen, & Voigt, 1991; Sinkjær, Andersen, Nielsen, & Hansen, 1999; Zuur et al., 2010). Amplitude of muscle activity prior to initial contact was determined from onset to initial contact. Amplitude of muscle activity following initial contact was evaluated in consecutive 30 ms epochs after initial contact (IC – 30 ms; 30 – 60 ms; 60 – 90 ms; 90 – 120 ms) and from peak vGRF to toe off (Figure 5.2) (Hobara et al., 2008; Horita, Komi, Nicol, & Kyröläinen, 1996; Kramer, Ritzmann, Gruber, & Gollhofer, 2012; Lloyd et al., 2012; Oliver & Smith, 2010; Ritzmann, Kramer, Gollhofer, & Taube, 2013). Amplitude normalisation was not performed as comparison was made between time periods for each participant (within-subject design).

Muscle activity from onset to initial contact was termed the feedforward phase and is of a central (supra-spinal) origin (Hobara et al., 2007; Horita et al., 1996; Zuur et al., 2010). The muscle activity in the 30 ms immediately following initial contact has been reported to be an extension of the feedforward activity (Hobara et al., 2007; Voigt, Dyhre-Poulsen, & Simonsen, 1998) and is commonly termed background activity. Muscle activity from 30 to 60 ms following initial contact has been reported to represent the short-latency stretch reflex (M1), which is a spinal-reflex and commonly observed in tasks such as walking, jumping and hopping (Dyhre-Poulsen et al., 1991; Horita et al., 1996; Ritzmann et al., 2013; Sinkjær et al., 1999; Zuur et al., 2010). Muscle activity from 60 to 90 ms following initial contact has been labelled the long-latency reflex (M2) and represents activity modulated by both spinal and supra-spinal reflexes (Horita et al., 1996; Sinkjær et al., 1999). Muscle activity from 90 to 120 ms following initial contact has been labelled a second long-latency reflex or functional stretch-reflex (M3) and represents activity of a predominantly supra-spinal origin (Melvill-Jones & Watt, 1971; Sinkjær et al., 1999). Muscle activity from peak vGRF to toe off was termed the propulsion phase and may represent concentric activity of the muscle occurring during the hop cycle that is of a central origin.

Muscle co-activation ratio was calculated for each phase following initial contact (background, M1, M2 and M3). Co-activation at the ankle (SOL:TA) and knee joints (VL:BF) were determined as the quotient of agonist and antagonist muscle MAV (Frey-Law & Avin, 2013; Lloyd et al., 2012; Padua et al., 2006).

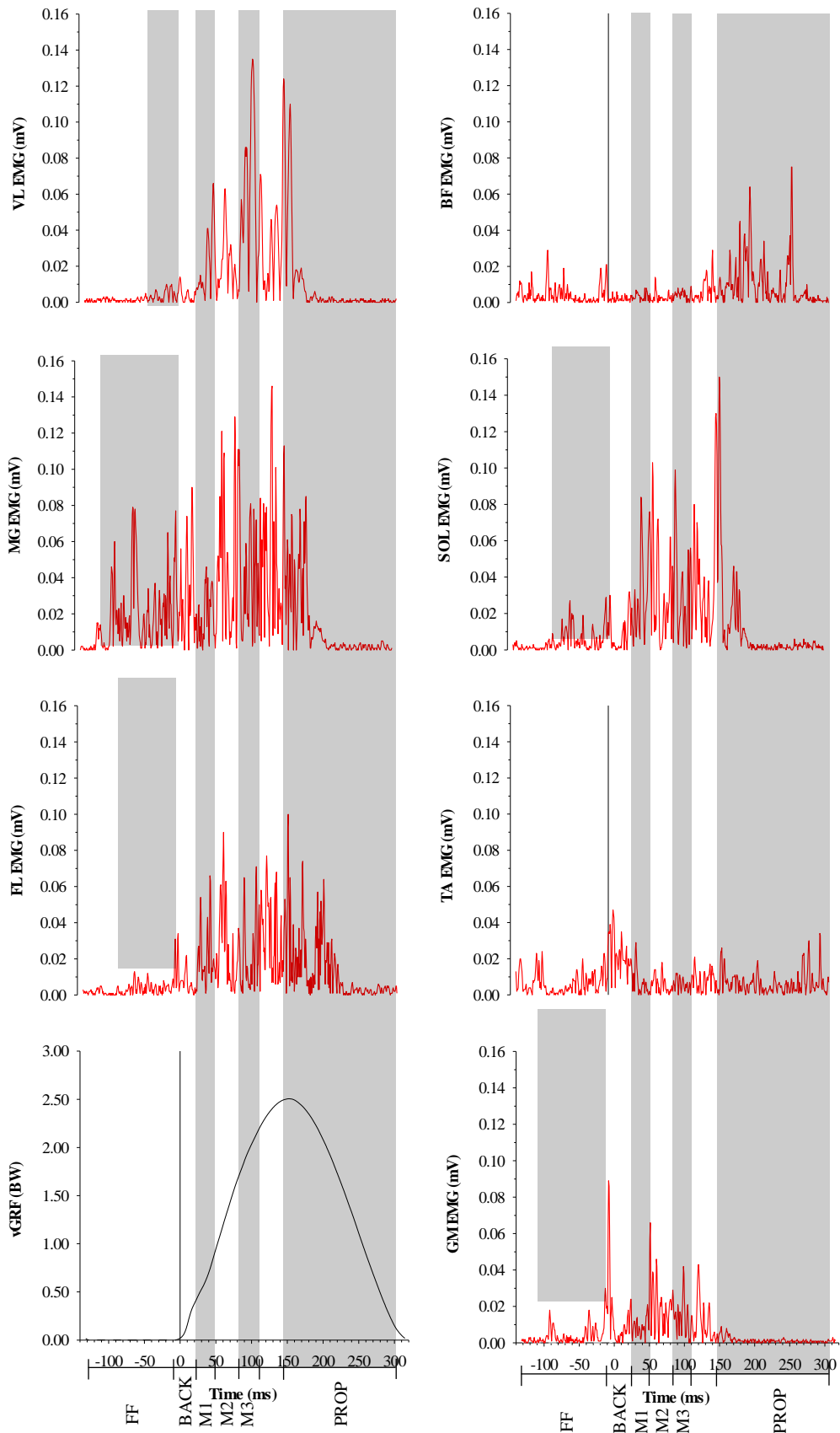


Figure 5.2 Vertical ground reaction force (vGRF) and filtered (band-pass filtered 50 to 500 Hz) and rectified surface electromyography (mV) signals for a single hop cycle for each lower limb muscle (GM, VL, BF, MG, SOL and TA). The phases of activity include the feedforward (FF), background (BACK), short-latency stretch reflex (M1), long-latency stretch reflex (M2), second long-latency stretch reflex (M3) and propulsion (PROP) phases.

5.2.8 Statistical analyses

A one-way (time period) repeated measures ANOVA (0, 20, 40, 60, 80 and 100% time periods) was performed to compare each derived variable at each time period during the exhaustive single-leg hopping trial (SPSS, Version 22). Mauchly's test of sphericity was performed and if violated ($p < 0.05$) a Greenhouse-Geisser procedure used. Multiple pairwise comparisons were made between the 0% time period and all subsequent time periods with a Bonferroni correction made to reduce the risk of a type 1 error. Alpha levels were set *a priori* with significance accepted at $p < 0.05$.

5.3 Results

5.3.1 Performance output

A detailed description of performance output was described in Chapter 4. In summary, during the exhaustive hopping trial there was no significant change detected in performance output characteristics, including hopping frequency, hopping height and vertical stiffness (Table 4.1).

5.3.2 Neuromuscular characteristics

5.3.2.1 Temporal characteristics

There was a significant decrease in the duration of the feedforward phase of the MG and SOL muscles at the 40% ($p < 0.001$; $p = 0.018$), 60% ($p < 0.001$; $p < 0.001$), 80% ($p < 0.001$; $p < 0.001$) and 100% ($p < 0.001$; $p = 0.014$) time periods compared to the start. The duration significantly decreased for the feedforward phase of the FL muscle at the 80% ($p = 0.028$) and 100% ($p < 0.001$) time periods, and the GM muscle at 100% ($p = 0.003$) time period compared to the start. In contrast, there was

a significant increase in the duration of the feedforward phase for the VL muscle at the 60% ($p = 0.041$) time period (Table 5.1; Appendix N.1).

There was a significant decrease in the total duration of MG and SOL muscle activity at the 40% ($p = 0.004$; $p = 0.001$), 60% ($p < 0.001$; $p < 0.001$) and 80% ($p < 0.001$; $p = 0.035$) time periods compared to the start. Total duration of muscle activity significantly decreased for the FL muscle at the 40% ($p = 0.001$) time period, and the FL and TA muscles at the 60% ($p < 0.001$; $p = 0.017$), 80% ($p = 0.004$; $p = 0.023$) and 100% ($p = 0.007$; $p = 0.041$) time periods compared to the start. In contrast, there was a significant increase in the total duration of VL muscle activity at the 80% ($p = 0.047$) and 100% ($p = 0.029$) time periods compared to the start (Table 5.1; Appendix N.2).

Table 5.1 Temporal characteristics (mean \pm SD) for the muscles acting at the hip, knee and ankle joints at all time periods.

	0%	20%	40%	60%	80%	100%
Feedforward						
Duration (ms)						
GM	80 \pm 20	78 \pm 27	77 \pm 24	76 \pm 21	77 \pm 25	67 \pm 21*
VL	50 \pm 28	54 \pm 23	54 \pm 24	58 \pm 24*	58 \pm 25	58 \pm 24
MG	97 \pm 18	91 \pm 21	84 \pm 22*	78 \pm 21*	72 \pm 18*	72 \pm 18*
SOL	80 \pm 25	75 \pm 23	70 \pm 24*	66 \pm 23*	66 \pm 23*	67 \pm 23*
FL	81 \pm 36	68 \pm 45	72 \pm 36	68 \pm 31	67 \pm 34*	60 \pm 31*
Total						
Duration (ms)						
GM	302 \pm 29	307 \pm 27	306 \pm 29	307 \pm 27	318 \pm 30	314 \pm 28
VL	275 \pm 37	272 \pm 37	268 \pm 35	278 \pm 38	286 \pm 38*	289 \pm 39*
MG	340 \pm 21	335 \pm 23	324 \pm 26*	319 \pm 25*	322 \pm 25*	329 \pm 24
SOL	328 \pm 29	322 \pm 28	310 \pm 31*	309 \pm 27*	315 \pm 26*	324 \pm 30
FL	338 \pm 28	322 \pm 38	319 \pm 31*	316 \pm 31*	321 \pm 30*	318 \pm 30*
TA	172 \pm 40	159 \pm 31	164 \pm 36	149 \pm 24*	152 \pm 30*	149 \pm 33*

* = $p < 0.05$ compared to the 0% time period

5.3.2.2 *Amplitude characteristics*

During the feedforward phase, there was a significant increase in MAV of the SOL muscle at the 20% ($p = 0.035$) time period, and SOL and FL muscles at the 40% ($p = 0.010$; $p = 0.003$), 60% ($p < 0.001$; $p = 0.004$), 80% ($p < 0.001$; $p = 0.046$) and 100% ($p < 0.001$; $p = 0.004$) time periods compared to the start of the trial. There was a significant increase in MAV of the VL muscle at the 80% ($p = 0.001$) and 100% ($p = 0.001$) time periods compared to the start of the trial (Table 5.2; Appendix N.3).

Compared to the start of the trial, during the background phase there was a significant increase in MAV of the SOL muscle at the 40% ($p = 0.013$), 60% ($p < 0.001$), 80% ($p < 0.001$) and 100% ($p < 0.001$) time periods. The MAV significantly increased for the BF muscle at the 40% ($p = 0.002$) time period, and BF and FL muscles at the 60% ($p = 0.042$; $p = 0.006$) time period compared to the start of the trial. Further, the MAV of the VL muscle significantly increased at the 80% ($p = 0.042$) and 100% ($p = 0.011$) time periods compared to the start of the trial. In contrast, there was a significant decrease in MAV of the TA muscle at the 40% ($p = 0.023$), 60% ($p = 0.031$) and 80% ($p = 0.036$) time periods compared to the start of the trial (Table 5.2; Appendix N.4).

For the M1 phase, there was a significant increase in MAV of the SOL muscle at all time periods ($p < 0.001$) compared to the start of the trial. The MAV of the BF, VL and MG muscles significantly increased at the 40% ($p = 0.001$; $p = 0.002$; $p = 0.049$), 60% ($p = 0.001$; $p = 0.001$; $p < 0.001$) and 80% ($p = 0.001$; $p < 0.001$; $p = 0.015$) time periods, and BF and VL muscles at the 100% ($p = 0.005$; $p < 0.001$) time period compared to the start of the trial. Compared to the start of the trial there was a significant increase in MAV of the FL muscle at the 60% ($p = 0.018$) time period and the GM muscle at the 80% ($p = 0.018$) and 100% ($p = 0.010$) time periods (Table 5.2; Appendix N.5).

For the duration of the M2 phase, there was a significant increase in MAV of the SOL muscle at the 20% ($p = 0.023$) time period and all other time periods ($p < 0.001$) compared to the start of the trial. The MAV of the GM, BF and VL muscles significantly increased at the 40% ($p < 0.001$; $p = 0.001$; $p = 0.002$), 60% ($p = 0.003$; $p < 0.001$; $p = 0.001$), 80% ($p = 0.006$; $p = 0.014$; $p = 0.002$) and 100% ($p = 0.004$; $p = 0.009$; $p < 0.001$) time periods compared to the start of the trial. There was a

significant increase in MAV of the FL muscle at the 40% ($p < 0.001$) and 60% ($p = 0.035$) time periods, and MG muscle at the 60% ($p = 0.027$) and 80% ($p = 0.048$) time periods compared to the start of the trial (Table 5.2; Appendix N.6).

During the M3 phase, there was a significant increase in MAV of the SOL muscle at the 20% ($p = 0.015$) time period, and SOL and BF muscles at the 40% ($p = 0.001$; $p = 0.002$), 60% ($p < 0.001$; $p = 0.001$), 80% ($p < 0.001$; $p = 0.001$) and 100% ($p = 0.011$; $p = 0.042$) time periods compared to the start of the trial. The MAV of the VL muscle significantly increased at the 20% ($p = 0.029$), 60% ($p = 0.014$), 80% ($p = 0.004$) and 100% ($p < 0.001$) time periods compared to the start of the trial. Further, there was a significant increase in MAV of the GM muscle at the 60% ($p = 0.042$) and 100% ($p = 0.020$) time periods compared to the start of the trial (Table 5.2; Appendix N.7).

Compared to the start of the trial, during the propulsion phase there was a significant increase in MAV of the TA muscle at the 40% ($p = 0.015$) and 60% ($p = 0.001$) time periods, and TA and VL muscles at the 80% ($p < 0.001$; $p = 0.048$) and 100% ($p < 0.001$; $p = 0.004$) time periods. The MAV of the GM, MG and SOL muscles significantly increased at the 100% ($p = 0.003$; $p = 0.002$; $p < 0.017$) time period compared to the start of the trial (Table 5.2; Appendix N.8).

The co-activation ratio of the SOL:TA muscle pair significantly increased during the M1 phase at all time periods ($p \leq 0.004$) compared to the start of the trial. Further, there was a significant increase in co-activation of the SOL:TA muscle pair during the background phase at the 60% ($p < 0.001$), 80% ($p < 0.001$) and 100% ($p = 0.007$) time periods, and during the M2 phase at the 40% ($p = 0.035$) time period compared to the start of the trial (Figure 5.3; Appendix N.9).

Table 5.2 Mean amplitude value (MAV) (mean \pm SD) for the muscles acting at the hip, knee and ankle joints at all time periods. The MAV of each muscle was normalised to the value at the start of the trial (0%) for the table only.

	0%	20%	40%	60%	80%	100%
Feedforward						
GM	100	102 \pm 26	111 \pm 30	113 \pm 30	118 \pm 39	127 \pm 38
VL	100	105 \pm 24	119 \pm 30	129 \pm 41	163 \pm 74*	192 \pm 101*
MG	100	107 \pm 13	99 \pm 15	95 \pm 18	97 \pm 22	97 \pm 25
SOL	100	114 \pm 29*	133 \pm 56*	160 \pm 64*	191 \pm 95*	230 \pm 131*
FL	100	113 \pm 17	126 \pm 30*	127 \pm 29*	133 \pm 44*	147 \pm 52*
Background						
GM	100	108 \pm 25	113 \pm 28	118 \pm 27	121 \pm 38	124 \pm 43
BF	100	107 \pm 45	131 \pm 18*	132 \pm 52*	141 \pm 70	139 \pm 65
VL	100	112 \pm 36	125 \pm 36	130 \pm 49	151 \pm 85*	169 \pm 91*
MG	100	105 \pm 18	97 \pm 17	96 \pm 18	99 \pm 24	99 \pm 23
SOL	100	119 \pm 39	136 \pm 52*	174 \pm 92*	185 \pm 93*	201 \pm 114*
FL	100	120 \pm 40	128 \pm 42	136 \pm 43*	128 \pm 51	140 \pm 60
TA	100	90 \pm 33	88 \pm 31*	85 \pm 34*	90 \pm 45*	100 \pm 54
M1						
GM	100	110 \pm 25	120 \pm 36	122 \pm 39	127 \pm 41*	132 \pm 47*
BF	100	114 \pm 36	143 \pm 49*	147 \pm 50*	151 \pm 62*	161 \pm 77*
VL	100	114 \pm 28	137 \pm 43*	153 \pm 58*	168 \pm 70*	185 \pm 96*
MG	100	112 \pm 23	113 \pm 21*	120 \pm 25*	120 \pm 28*	118 \pm 30
SOL	100	124 \pm 23*	146 \pm 36*	154 \pm 35*	162 \pm 43*	167 \pm 46*
FL	100	113 \pm 26	129 \pm 41	131 \pm 38*	135 \pm 43	135 \pm 46
TA	100	97 \pm 25	105 \pm 37	111 \pm 43	117 \pm 55	120 \pm 51

Table 5.2 continued next page.

Table 5.2 continued.

	0%	20%	40%	60%	80%	100%
M2						
GM	100	114 ± 23	125 ± 24*	133 ± 29*	141 ± 34*	150 ± 40*
BF	100	126 ± 47	152 ± 61*	155 ± 67*	156 ± 80*	163 ± 85*
VL	100	121 ± 46	141 ± 46*	152 ± 61*	154 ± 58*	175 ± 86*
MG	100	109 ± 23	112 ± 25	116 ± 27*	118 ± 32*	124 ± 44
SOL	100	113 ± 22*	132 ± 30*	136 ± 33*	144 ± 38*	144 ± 45*
FL	100	113 ± 29	126 ± 31*	135 ± 45*	125 ± 41	133 ± 50
TA	100	94 ± 33	105 ± 36	114 ± 45	115 ± 46	122 ± 49
M3						
GM	100	113 ± 26	115 ± 28	126 ± 35*	135 ± 40*	151 ± 50*
BF	100	121 ± 39	140 ± 45*	155 ± 63*	143 ± 49*	150 ± 66*
VL	100	121 ± 31*	129 ± 37	145 ± 49*	153 ± 57*	169 ± 67*
MG	100	105 ± 18	108 ± 20	110 ± 23	110 ± 20	108 ± 28
SOL	100	112 ± 19*	122 ± 25*	128 ± 26*	130 ± 33*	131 ± 40*
FL	100	114 ± 31	123 ± 33	126 ± 42	112 ± 36	110 ± 39
TA	100	96 ± 25	107 ± 28	116 ± 31	119 ± 39	122 ± 43
Propulsion						
GM	100	102 ± 35	109 ± 48	124 ± 48	147 ± 69	174 ± 76*
BF	100	99 ± 24	104 ± 28	108 ± 30	113 ± 46	131 ± 65
VL	100	95 ± 31	101 ± 41	120 ± 56	156 ± 73*	192 ± 107*
MG	100	110 ± 20	108 ± 24	107 ± 25	115 ± 28	128 ± 41*
SOL	100	100 ± 17	102 ± 21	104 ± 20	113 ± 29	121 ± 33*
FL	100	103 ± 20	105 ± 21	105 ± 23	105 ± 30	111 ± 38
TA	100	107 ± 22	121 ± 32*	129 ± 32*	130 ± 35*	144 ± 48*

* = $p < 0.05$ compared to the 0% time period, M1 = short-latency stretch reflex, M2 = long-latency reflex, M3 = second long-latency reflex

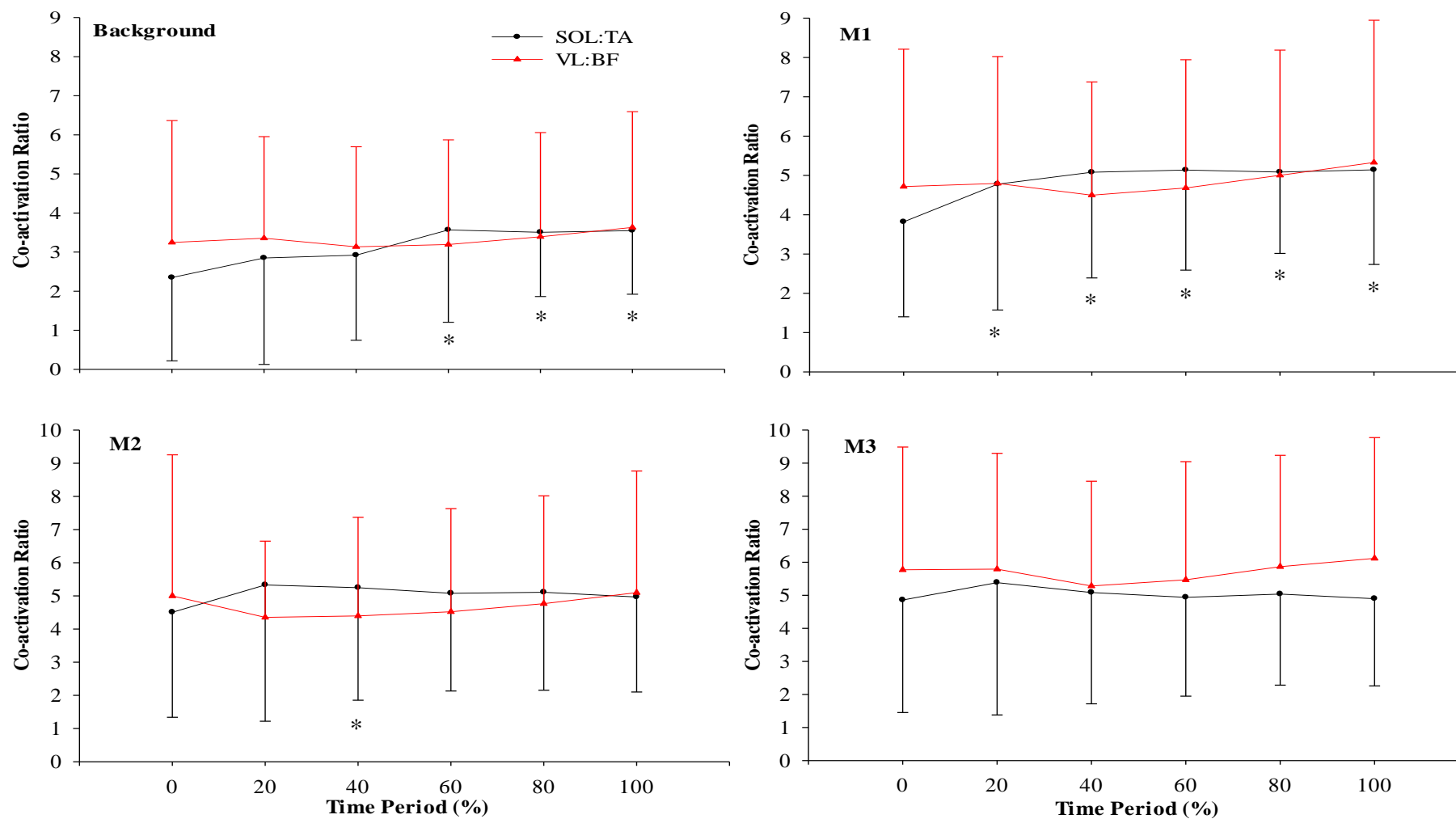


Figure 5.3 Co-activation ratio of SOL:TA and VL:BF for each time period. * = $p < 0.05$ compared to 0% time period for SOL:TA.

5.4 Discussion

This study evaluated changes in neuromuscular activity when performance output was maintained as muscle fatigue progressed during a single-leg hopping task. The key findings of the current study were: 1) increased SOL muscle activity without a concomitant increase in MG muscle activity; 2) increased VL muscle activity that occurred from the mid-late stages of the exhaustive trial; and 3) increased feedforward and feedback muscle activity as local muscle fatigue progressed. These findings suggest that as single-leg hopping progresses to exhaustion, there was a differential change within the triceps surae muscle group, which acted at the ankle joint and has been shown to be the site for modulation when performing repeated hopping (Lamontagne & Kennedy, 2013). Further, this study observed there was a specific shift in neuromuscular activation patterns of lower limb muscles when the overall performance including leg mechanical characteristics remained unchanged.

Consistent increases in the MAV of the SOL muscle during all phases throughout the progression of the task were observed. In contrast, the MAV of the MG muscle only increased during the M1 and M2 phases at the 40% to 80% time periods. Duration of activity during the feedforward phase and total duration of activity for SOL and MG muscles decreased initially from the start to the 40% time period, then remained relatively unchanged until the cessation of the fatigue protocol. It was evident that as the task progressed and there was an increasing level of fatigue, there was a consistent increase in activation of the SOL muscle but not the MG muscle. Whilst an increased amplitude does indicate activation of a greater number of muscle fibres (Gandevia, 2001), it may be speculated that the functional role of the SOL and MG were different to maintain leg mechanical characteristics during hopping.

Considering the repetitive and rhythmical hopping task is likely to have led to an increase in local muscle fatigue (Komi, 2000; Nicol et al., 2006), there may have been a greater reliance on specific muscles that were well suited to generate appropriate tensile force to maintain the specific task performance. During single-leg the SOL may have been more suited to the exhaustive task than the MG due to a greater proportion of slow-twitch oxidative fibres (Gollnick, Sjödín, Karlsson, Jansson, & Saltin, 1974) and a different angle of pennation (Kawakami, Ichinose, & Fukunaga, 1998). These characteristics of the SOL may improve the sensitivity of

the stretch-reflex (Maton & Pellec, 2001) as well as decrease its fatigability when compared to the MG muscle.

It is also plausible that there were differences between the SOL and MG in supraspinal level mechanisms in the response to repetitive loading. This notion is supported by the observation of consistent increases in activation of the SOL during the feedforward phase when there were no such changes observed for the MG. Further, during loading the SOL had a consistent increase in levels of activation, which were mediated at the spinal level (M1 and M2 phases) (Dyhre-Poulsen et al., 1991; Horita et al., 1996; Sinkjær et al., 1999), and those that may have been modulated by central mechanisms (background and M3 phases) (Hobara et al., 2007; Melvill-Jones & Watt, 1971; Sinkjær et al., 1999; Voigt et al., 1998). The findings in the current study contrast those of previous studies that have reported decreases in both central and spinal-reflex muscle activity following 60 s of maximal height hopping (Moritani et al., 1990), an exhaustive sledge rebound trial (Kuitunen et al., 2007) and submaximal double-leg hopping to exhaustion (Kuitunen et al., 2002). Nevertheless, these previous studies reported significant changes in performance output such as a decrease in flight height (Moritani et al., 1990), leg stiffness (Kuitunen et al., 2007) and take off velocity (Kuitunen et al., 2002). Thus, it is probable that the observed decreases in muscle activity in these studies were due to the changes in spatio-temporal and leg mechanical characteristics rather than fatigue only (Santello, 2005; Santello & McDonagh, 1998). The current study was able to avoid the possible confounding factors of a change in spatio-temporal and leg mechanical characteristics by controlling performance output. Hence, the differential activation patterns between SOL and MG muscles were able to be identified.

Increased levels of activation of the VL were consistent with changes in SOL muscle activation during the M1 and M2 phases, occurring from the 20% to 40% time periods' onwards. This suggests that the VL stretch reflex was modulated similarly to the SOL muscle. In contrast, a significant increase in VL activity during the centrally driven feedforward and background phases (80% time period) occurred later in the trial compared to the SOL muscle (20% to 40% time periods). Thus, there may be an increase in central drive to the VL muscle as the task progressed, towards the latter stages of the trial.

Repeated submaximal hopping has been reported to be modulated by ankle function (Bonnard et al., 1994; Dal Pupo et al., 2013; Farley & Morgenroth, 1999; Hobara et al., 2008; Lamontagne & Kennedy, 2013). Although, it is plausible that there were changes in muscles acting about the knee or hip as a strategy to maintain the desired performance output later in the trial when there was a greater progression of fatigue. Fatigue is suggested to be a progressive process in which the force generating capacity of the muscles begins to decline at almost the onset of activity (Bigland-Ritchie & Woods, 1984; Gandevia, 2001). Therefore, compensation within the predominant joint may be the preferred strategy during the early stages of a repetitive task when muscle fatigue would be much less developed than later in the trial. However, increased central drive of the VL muscle may be a strategy whereby there is a greater relative contribution from potentially less fatigued musculature towards the latter stages of an exhaustive task.

It is probable that the activity recorded during the later period of the loading phase (M3 – functional stretch-reflex) was in preparation for propulsion as the muscle activity will precede any observed movement changes due to electromechanical delay (Melvill-Jones & Watt, 1971; Sinkjær et al., 1999). The progressively increased levels of activation of the VL and SOL would have been due to the demand of having to maintain the height of hopping throughout the task. Therefore, the synergistic action of the ankle plantarflexor and knee extensor allowed maintenance of extension of the joints into propulsion which was synchronous and probable that the changes in activity were a centrally mediated mechanism.

The current study supports the importance of central and spinal-reflex muscle activity for the regulation of lower limb stiffness in a stereotypical task. Greater levels of lower limb stiffness have been shown to require a lower energy cost during running (Dalleau, Belli, Bourdin, & Lacour, 1998), and were likely due to alterations of how the SSC was modulated and contributed to the generation of positive joint work. Therefore, the overall increase in muscle activity during the feedforward and feedback phases in the current study may have contributed to the ability to maintain stiffness and sustain a minimalistic energy cost as muscle fatigue progressed and muscle function deteriorated.

In addition to the significant increase in muscle activity of the prime movers (SOL and VL), there was a significant decrease in muscle activity of the antagonist muscle such as the TA muscle during the background phase. This finding is in agreement with previous research (Padua et al., 2006) that demonstrated a significant decrease in TA activity during double-leg hopping following a fatiguing squat task. The relative inhibition of the TA muscle and increase in SOL muscle activity led to a finding of increased co-activation at the ankle joint during the initial part of the loading phase as the task progressed. This had the potential to modulate ankle stiffness as the task progressed and optimise the functional role of the ankle as fatigue increased.

In contrast, co-activation at the knee remained unchanged due to a concomitant increase in BF activity during the loading phase. The increase in BF activity may be due to it being a biarticular muscle in which a stretch-reflex is induced when hip flexion occurs during loading. Muscle co-activation is believed to contribute to joint stability (Frey-Law & Avin, 2013), with increased co-activation suggested to improve the transfer of stored elastic energy to the concentric phase (Padua et al., 2006). A significant increase in muscle co-activation at the ankle and not the knee joint further supports adaptation at the predominant joint as the preferred strategy when maintaining performance during a submaximal exhaustive task.

As the hopping task progressed there was a significant increase in muscle activity of the GM, VL and SOL muscles during the propulsion phase at the 80% and 100% time periods. Increased activity of the agonist muscles during the concentric phase of the task may have assisted the generation of positive mechanical work during the propulsion phase to achieve the required hop height. This is a plausible consequence to the likely decreased tolerance to repeated stretch loads that occur during the fatiguing SSC task, which has been shown to lead to a reduced ability to utilise stored elastic energy during the concentric phase (Gollhofer, Komi, Miyashita, & Aura, 1987; Komi, 2000). Producing positive work during the concentric phase has a much greater metabolic energy cost than producing an equal magnitude of negative work during the eccentric phase (Moritani, Muramatsu, & Muro, 1987; Pasquet, Carpentier, Duchateau, & Hainaut, 2000). It is possible increased muscle activity during the propulsion phase was a compensation for a decreased contribution from elastic energy, but may further augment muscle fatigue (Komi, 2000; Kuitunen et al.,

2007). Therefore, no changes in muscle activity during the propulsion phase until the latter stages of the exhaustive trial is hypothesised to be a strategy to ensure a minimisation of energy cost throughout the exhaustive task. The increase in muscle activity during the propulsion phase that occurred towards the end of the trial may act as a “final effort” by the central nervous system to maintain the performance output immediately prior to exercise cessation.

5.5 Conclusion

Differential increases in centrally mediated and spinal-reflex activity demonstrate the specific compensatory strategies that may be adopted during performance of stereotypical, submaximal height hopping to exhaustion. Importantly, incremental and consistent increases were observed in the VL and SOL muscles demonstrating a centrally mediated synergy between segments. The observations in this study highlight that neuromuscular changes are sensitive to the task which in this case was being able to maintain stereotypical, submaximal effort hopping as fatigue progressed.


CHAPTER SIX


A comparison of neuromechanical changes between men and women during a repetitive loading task to exhaustion

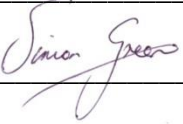
I, Kurt Mudie, hereby declare that I was the principal researcher of all work included in this chapter. A statement from the co-authors confirming contribution of the PhD candidate to the work is presented below.

As co-authors of the chapter ‘A comparison of neuromechanical changes between men and women during a repetitive loading task to exhaustion’, we confirm that Kurt Mudie made the following contributions:

- Conception and design of the research question
- Data collection, processing and analysis
- Statistical analyses and interpretation of findings
- Writing and review of the chapter

Peter J Clothier:  Date: 6/01/17

Amitabh Gupta:  Date: 04/01/17

Simon Green:  Date: 16/01/2017

6 Introduction

Differences between men and women are often characterised in terms of strength, flexibility, endurance and cognitive abilities (Bruton et al., 2013; Jiménez-Jiménez et al., 2011; Thomas & French, 1985). In addition, there have been reported differences in the rate of injuries between sexes. For example, women have been reported to have a disproportionately greater rate of anterior cruciate ligament (ACL) rupture compared to men (Agel, Arendt, & Bershadsky, 2005; Arendt, Agel, & Dick, 1999). Although differences in movement outcome have been reported during landing (Bruton et al., 2013), it still remains unclear whether there is a sex-based bias in movement strategies that have been suggested to be precipitant to ACL rupture (Breen, 2012) and patellofemoral pain syndrome (Heiderscheit et al., 2002). A large proportion of research exploring differences between sexes in biomechanical and neuromuscular characteristics has examined tasks such as running, single/double leg drop landings and cut tasks during non-fatiguing conditions (Bruton et al., 2013). It remains unclear whether there is a difference in how fatigue affects lower limb neuromechanical characteristics differently between the sexes.

Women have been shown to have lower levels of joint coupling variability than men during dynamic tasks such as an unanticipated cut manoeuvre in soccer players (Pollard et al., 2005) and hip, knee and ankle joint variability during treadmill running at 12 km·h⁻¹ (Barrett et al., 2008). It has been suggested that the lower variability in women may limit their ability to adapt to changes in the task or the environment (Barrett et al., 2008; Pollard et al., 2005) and may expose the musculoskeletal system to greater localised mechanical stress during repetitive tasks (Bartlett et al., 2007; Hamill et al., 2012; Hamill et al., 1999; Preatoni et al., 2013). Although these studies have examined differences between the sexes during rapid and repeated loading tasks, it is not known whether these differences are evident during tasks to exhaustion.

There are a number of potential differences in neuromechanical characteristics between men and women. It has been suggested that women adopt a “quadriceps dominant” strategy during landing (Griffin et al., 2006; Shultz et al., 2009). This has been observed as greater rectus femoris muscle activity prior to landing (Nagano et al., 2007) and vastus medialis activity during single-leg landing from a vertical jump

(Urabe et al., 2005) in women compared to men. During a double-leg drop-jump, women have displayed greater MAV of the quadriceps and hamstring muscles prior to and during loading (Shultz et al., 2009). However, a quadriceps dominant strategy has not been a consistent finding during fatiguing tasks, with no differences reported between sexes when landing from a double-leg drop jump (Gehring et al., 2009) or during double-leg hopping (Padua et al., 2006; Padua, Garcia, Arnold, & Granata, 2005). It is plausible that the studies by Gehring et al. (2009) and Padua et al. (2006) did not detect a statistically significant difference between sexes due to relatively small sample sizes of 26 and 21 participants, respectively. Additionally, the fatiguing protocol utilised an exhaustive leg-press exercise (Gehring et al., 2009) and body weight squats (Padua et al., 2006), which required different movement strategies than the tasks which were used to evaluate the changes in neuromechanical characteristics. Thus, changes in neuromuscular activity were more likely specific to the fatiguing task and not representative of changes during a repetitive loading or landing task. Of interest in this current study was to determine if there were differences between sexes during a continuous repetitive loading task that may better represent the demands in common sporting activities such as running and jumping that employ the SSC muscle action.

Differences in neuromechanical characteristics between sexes during rapid and repeated loading tasks exist (Barrett et al., 2008; Bruton et al., 2013; Pollard et al., 2005), yet it remains unknown how fatigue influences these sex differences in neuromechanical characteristics. The purpose of this study was to evaluate the effect of sex on lower limb joint coupling variability and neuromuscular characteristics during on-the-spot, single-leg hopping to exhaustion. It was hypothesised that there would be a greater increase in joint coupling variability and neuromuscular characteristics in women compared to men during on-the-spot, single-leg hopping to exhaustion.

6.2 Methods

Data which was analysed to test the hypothesis in this study had been collected from the same participants reported in Chapter 4. Methods common to the current Chapter and Chapters 3, 4 & 5 are presented here in brief.

6.2.1 Participants

Participants comprised the same sample group described in Chapters 4 and 5. Specifically, forty healthy recreationally active men ($n = 20$; 23 ± 3 years of age; 1.8 ± 0.1 m height; 76.4 ± 7.6 kg body mass) and women ($n = 20$; 22 ± 3 years of age; 1.7 ± 0.1 m height; 61.3 ± 7.7 kg body mass) volunteered to participate in this study. Participants met the inclusion criteria described in Chapter 3 and ethical approval was granted by the University of Western Sydney Human Research Ethics Committee (Appendix F).

6.2.2 Instrumentation

The instrumentation and software used to collect kinetic, kinematic and sEMG data synchronously were described in Chapters 3, 4 and 5. Briefly, kinetic, kinematic and EMG data were collected synchronously during single-leg, on-the-spot hopping to volitional exhaustion. The instrumentation used to collect the kinetic and kinematic data were described in Chapter 3 (Section 3.2.2) and EMG data in Chapter 5 (Section 5.2.2). All data were time synchronised, recorded (First Principles, Version 1.2.4) and post-processed offline following data collection (Visual 3D, C-Motion, Version 4, Germantown, MD).

6.2.3 Participant preparation

Participant preparation was described in Chapters 3 and 5. Briefly, following the measurement of height and body mass, a warm up and familiarisation period was completed. Marker clusters and individual markers were attached to the trunk and lower limbs (Section 3.2.3.1; Appendix J). A seven segment model was used to model the trunk, pelvis, thigh, shank, hindfoot, forefoot and hallux of the hopping leg (Appendix K). Skin preparation was consistent with SENIAM guidelines and sEMG electrodes were adhered over the GM, BF, VL, MG, SOL, FL and TA muscles of the hopping leg (Section 5.2.3; Appendix M).

6.2.4 Controlling task performance

An audible metronome was fixed at 2.2 Hz (132 beats per minute) (Farley et al., 1991; Hobara, Kobayashi, et al., 2013) and the target hop height was fixed for each participant as their preferred hopping height at 2.2 Hz (Section 4.2.4). Tactile

feedback was provided by a custom build apparatus to allow each person to hop to their target hopping height.

6.2.5 Testing protocol

A static calibration trial was recorded then each participant completed a single effort of on-the-spot, single-leg hopping to volitional exhaustion at 2.2 Hz to a target hop height. Five continuous maximal effort single-leg hops were performed three minutes before (PRE) and 10 s after (POST) the exhaustive single-leg hopping trial.

6.2.6 Data processing

All recorded data were exported (First Principles, Version 1.2.4) to be later analysed using Visual 3D (Version 4). Force platform data were filtered using a fourth order bidirectional low-pass Butterworth filter with a 50 Hz cut-off (Gupta et al., 2016; Gupta et al., 2014). Recorded marker trajectories were interpolated using a spline filter for up to a maximum gap of 10 frames and filtered using a fourth order bidirectional low-pass Butterworth filter with an 8 Hz cut-off frequency (Bobbert & Richard Casius, 2011; Hobara et al., 2011; Hobara et al., 2008). A 13 ms sEMG electromechanical delay was corrected for and the sEMG signals were filtered with a fourth order, bidirectional band-pass filter (50 to 500 Hz) and full wave rectified (Gupta et al., 2016; Gupta et al., 2014).

6.2.7 Data analysis

A single hop cycle was defined as a complete flight phase and subsequent contact phase determined from the vGRF trace (Figure 3.6). Dependent variables calculated for each hop cycle during the exhaustive trial included hopping frequency (Hz), ΔCOM_f (cm), contact time (s), K_{vert} ($\text{N}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$), joint kinetics and kinematics, lower limb joint coupling variability and temporal and amplitude muscle activation characteristics derived from sEMG signals. Dependent variables were calculated as the mean of 10 consecutive hop cycles at time periods of 0, 20, 40, 60, 80 and 100% of the duration of the trial (Microsoft Office Excel, 2007 and MATLAB, 2012B 32-bit). To quantify the acute functional effects at volitional exhaustion, the single-hop cycle with the greatest hop height was chosen during the PRE and POST maximal effort hop trials.

6.2.7.1 Performance output

Vertical stiffness, ΔCOM_F and ΔCOM_L were determined by a segmental analysis method (Chapter 3) (Mudie et al., 2016).

6.2.7.2 Joint kinetics and kinematics

Hip, knee and ankle joint range of motion (ROM) was calculated in the sagittal plane during the loading phase. Joint stiffness was calculated as the ratio of change in joint moment and joint ROM during the loading phase, normalised to body mass (Visual 3D, Version 4). The direction of positive and negative motion was defined by the right-handed Cartesian local coordinate system (G. Wu et al., 2002).

6.2.7.3 Joint coupling variability

Joint coupling variability was calculated for knee flexion/extension - ankle flexion/extension (KxAx), hip flexion/extension - knee flexion/extension (HxKx) and knee flexion/extension - ankle eversion/inversion (KxAy) during the loading and propulsion phases using a modified vector coding technique (Section 4.2.7.2) (Mudie et al., 2016).

6.2.7.4 Neuromuscular characteristics

Briefly, EMG temporal and amplitude characteristics were determined as described in Chapter 5 (Section 5.2.7.2). Temporal characteristics for each hop cycle included the duration of the feedforward phase and total duration of muscle activity. The MAV of each muscle was determined for six predefined phases during each hop cycle, including the feedforward, background, M1, M2, M3 and propulsion phase (Figure 5.2). Amplitude normalisation of the MAV for each muscle was performed at each time period, relative to the 0% time period. This process allowed a comparison of sEMG signal between men and women. Muscle co-activation ratios were determined for the VL:BF and SOL:TA muscle pairs for the background, M1, M2 and M3 phases.

6.2.8 Statistical analyses

A two-tailed independent sample *t*-test was performed to compare the total duration of the exhaustive hopping trials between the sexes. The maximal height of hopping was compared from before (PRE) and after (POST) the exhaustive on-the-spot, single-leg hopping effort for men and women separately, using a two-tailed paired sample *t*-test. A two factor repeated measures ANOVA (sex with two levels (men and women); time period with six levels (0, 20, 40, 60, 80 and 100%)) was performed to determine differences for dependent variables including performance, joint kinetics and kinematics, joint coupling variability and neuromuscular characteristics. Mauchly's test of sphericity was performed and if violated ($p < 0.05$) a Greenhouse-Geisser procedure used. If there was a significant interaction or main effect, a *post-hoc* one-way ANOVA (time period) and/or independent samples *t*-tests (sex) were performed. If there was a significant main effect for time period, a one-way ANOVA with multiple pairwise comparisons were made between the 0% time period and all subsequent time periods for men and women, with a Bonferroni correction applied. If there was a significant main effect for sex, independent samples *t*-tests were performed to compare between men and women at each time point. Alpha levels were set *a priori* with significance accepted at $p < 0.05$. *Cohen's d* effect sizes were calculated with thresholds (0.2 = small, 0.6 = moderate, 1.2 = large and 2.0 = very large) used to qualitatively describe effect sizes (Hopkins, 2006). Increases and decreases in magnitude were indicated by positive and negative effect sizes, respectively.

6.3 Results

6.3.1 Fatigue and performance output

There was no significant difference observed in total duration of hopping (mean \pm SD) between men (79 ± 30 s) and women (78 ± 22 s; $p = 0.861$, $d = 0.056$) (Appendix O.1). There was a significant decrease in maximum height of hopping by 25% (PRE: 10.66 ± 2.99 ; POST: 8.04 ± 2.73 ; $p < 0.001$; $d = -0.88$) and 21% (PRE: 8.16 ± 2.04 ; POST: 6.47 ± 2.03 ; $p < 0.001$; $d = -0.83$) following the exhaustive trial of on-the-spot, single-leg hopping in men and women, respectively (Appendix O.2).

Performance output characteristics during the exhaustive trial, including hopping frequency, ΔCOM_f or K_{vert} were similar between men and women (Table 6.1; Appendix O.3). Further, there was no significant difference between men and women for normalised peak vGRF, and spatio-temporal characteristics including ΔCOM_L and the duration of the contact phase, loading phase, propulsion phase and flight phase (Table 6.2; Appendix O.3).

A main effect for time was observed for hopping frequency, K_{vert} , ΔCOM_L and duration of the contact, propulsion and flight phases. However, there was no significant change in hopping frequency, ΔCOM_f or K_{vert} between each time period for men. Women when compared to men demonstrated a small but significant decrease in hopping frequency between the 0% time period and the 20% time period (mean difference = -0.03 Hz, $p = 0.044$, $d = -0.836$), and K_{vert} between the 0% time period and the 80% (mean difference = $-14.97 \text{ N}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$, $p = 0.029$, $d = -0.628$) time period (Table 6.1; Appendix O.3). There were no significant changes between time periods in normalised peak vGRF and duration of the contact phase, loading phase and flight phase for either sex (Appendix O.4). No differences in the duration of the propulsion phase and ΔCOM_L between time periods were evident for men, whereas women demonstrated a significant increase in ΔCOM_L ($F = 9.507$, $p < 0.001$) as time progressed. Women demonstrated a small but significant increase in the duration of the propulsion phase between the 0% time period and the 80% (mean difference = 0.01 s, $p = 0.012$, $d = 0.737$) and 100% (mean difference = 0.01 s, $p = 0.002$, $d = 1.117$) time periods (Table 6.2).

Table 6.1 Performance output characteristics (mean \pm SD) for men and women at each time period during single-leg hopping to exhaustion.

	Sex	0%	20%	40%	60%	80%	100%
Hopping Frequency (Hz)	Men	2.21 \pm 0.06	2.21 \pm 0.03	2.22 \pm 0.04	2.23 \pm 0.04	2.22 \pm 0.05	2.23 \pm 0.05
	Women	2.25 \pm 0.04	2.22 \pm 0.03*	2.23 \pm 0.04	2.22 \pm 0.04	2.21 \pm 0.04	2.22 \pm 0.05
ΔCOM_f (cm)	Men	1.98 \pm 0.86	1.90 \pm 0.96	2.04 \pm 0.94	1.91 \pm 0.81	1.90 \pm 0.87	1.83 \pm 0.82
	Women	1.66 \pm 0.82	1.76 \pm 0.97	1.76 \pm 0.72	1.72 \pm 0.88	1.71 \pm 0.90	1.58 \pm 0.70
K_{vert} ($\text{N}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$)	Men	246.23 \pm 23.61	242.88 \pm 17.38	246.41 \pm 15.79	246.55 \pm 22.16	242.21 \pm 28.03	238.77 \pm 23.11
	Women	249.61 \pm 23.84	243.36 \pm 26.25	244.37 \pm 24.39	241.74 \pm 24.48	234.64 \pm 21.74*	235.59 \pm 20.58

* $p < 0.05$ compared to the 0% time period; ΔCOM_f = hopping height; K_{vert} = vertical stiffness

Table 6.2 Spatio-temporal characteristics and peak vertical ground reaction force (mean \pm SD) for men and women at each time period.

	Sex	0%	20%	40%	60%	80%	100%
Contact Phase (s)	Men	0.31 \pm 0.02	0.32 \pm 0.03	0.31 \pm 0.03	0.31 \pm 0.02	0.32 \pm 0.02	0.32 \pm 0.02
	Women	0.32 \pm 0.03	0.32 \pm 0.03	0.32 \pm 0.02	0.32 \pm 0.03	0.33 \pm 0.02	0.33 \pm 0.02
Loading Phase (s)	Men	0.15 \pm 0.01	0.15 \pm 0.01	0.15 \pm 0.01	0.14 \pm 0.02	0.15 \pm 0.02	0.14 \pm 0.02
	Women	0.15 \pm 0.01	0.15 \pm 0.01	0.15 \pm 0.01	0.15 \pm 0.01	0.15 \pm 0.01	0.15 \pm 0.01
Propulsive Phase (s)	Men	0.17 \pm 0.02	0.17 \pm 0.02	0.17 \pm 0.02	0.17 \pm 0.02	0.17 \pm 0.02	0.18 \pm 0.02
	Women	0.17 \pm 0.02	0.17 \pm 0.02	0.17 \pm 0.01	0.17 \pm 0.02	0.18 \pm 0.02*	0.18 \pm 0.02*
Flight Phase (s)	Men	0.14 \pm 0.02	0.14 \pm 0.03	0.14 \pm 0.03	0.13 \pm 0.02	0.13 \pm 0.02	0.13 \pm 0.02
	Women	0.13 \pm 0.02	0.13 \pm 0.03	0.13 \pm 0.02	0.13 \pm 0.02	0.12 \pm 0.02	0.12 \pm 0.02
ΔCOM_L (cm)	Men	10.23 \pm 0.95	10.29 \pm 0.85	10.31 \pm 0.73	10.32 \pm 0.87	10.46 \pm 0.88	10.50 \pm 0.88
	Women	9.86 \pm 0.70	10.06 \pm 0.88	10.14 \pm 0.75*	10.25 \pm 0.79*	10.47 \pm 0.82*	10.35 \pm 0.86*
Peak vGRF (N\cdotkg⁻¹)	Men	24.94 \pm 2.38	24.87 \pm 2.55	25.30 \pm 2.68	25.28 \pm 2.41	25.04 \pm 2.46	24.83 \pm 2.16
	Women	24.50 \pm 2.89	24.42 \pm 3.32	24.72 \pm 2.92	24.73 \pm 3.17	24.52 \pm 3.09	24.19 \pm 2.31

* $p < 0.05$ compared to the 0% time period, Δ COM_L = vertical displacement of the COM during loading, vGRF = vertical ground reaction force

6.3.2 Joint kinetics and kinematics

6.3.2.1 Differences between men and women

No between sex differences were observed for ankle, knee and hip joint ROM or stiffness during loading (Appendix O.5; Appendix O.6).

6.3.2.2 Differences between time periods for men and women

Main effects for time were observed for the knee and hip joint ROM during loading and ankle and knee joint stiffness during loading (Appendix O.5; Appendix O.6). *Post-hoc* one-way ANOVA results revealed both men and women demonstrated a significant decrease in knee joint ROM during loading ($F = 20.310, p < 0.001$; $F = 43.245, p < 0.001$), and a significant increase in knee joint stiffness during loading ($F = 8.404, p = 0.003$; $F = 10.718, p = 0.002$) as time progressed, respectively (Figure 6.1).

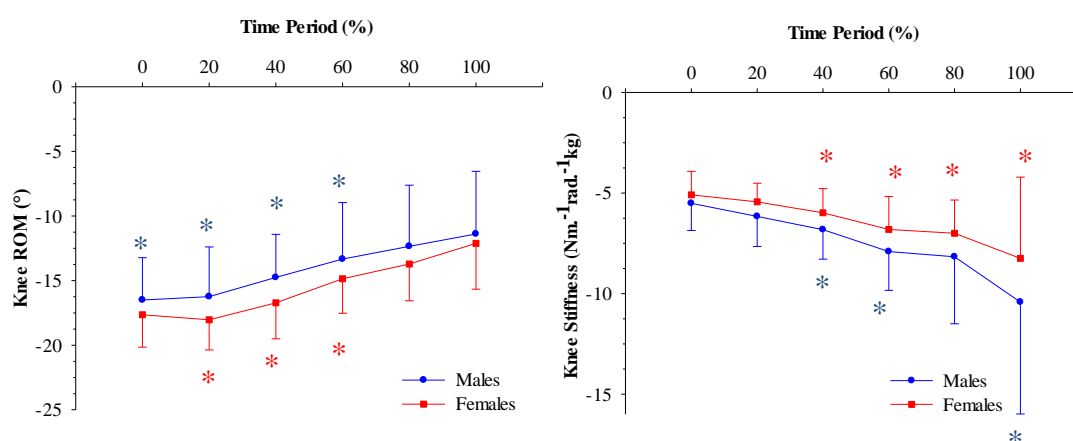


Figure 6.1 Knee joint ROM (°) and knee joint stiffness (Nm·rad⁻¹·kg⁻¹) (mean (SD)) during loading for men (blue) and women (red) at each time period. * = $p < 0.05$ compared to 0% time period for men (blue) and women (red).

6.3.3 Joint coupling variability

6.3.3.1 *Differences between men and women*

Main effects for sex were observed for the KxAx and KxAy couplings during the propulsion phase (Appendix O.7). Women demonstrated on *post-hoc* analysis a significantly lower KxAx compared to men at the 40% ($p = 0.033$, $d = 0.59$) and 60% ($p = 0.043$, $d = 0.67$) time periods and in KxAy at the 20% ($p = 0.038$, $d = 0.70$) time period during the propulsion phase (Figure 6.2).

6.3.3.2 *Differences between time periods for men and women*

Main effects for time were observed for the KxAx, HxKx and KxAy couplings during the loading and propulsion phase (Appendix O.7; Appendix O.8). *Post-hoc* one-way ANOVA results revealed both sexes demonstrated a significant increase in KxAx ($F = 8.855$, $p < 0.001$; $F = 10.934$, $p < 0.001$), HxKx ($F = 21.634$, $p < 0.001$; $F = 28.433$, $p < 0.001$) and KxAy ($F = 8.240$, $p < 0.001$; $F = 14.414$, $p < 0.001$) during the loading phase as time progressed, respectively (Figure 6.2).

Only women demonstrated a significant increase in KxAx during propulsion ($F = 6.635$, $p < 0.001$) as time progressed. Both sexes demonstrated a significant increase in HxKx during propulsion ($F = 7.512$, $p < 0.001$; $F = 18.678$, $p < 0.001$) as time progressed, respectively (Figure 6.2).

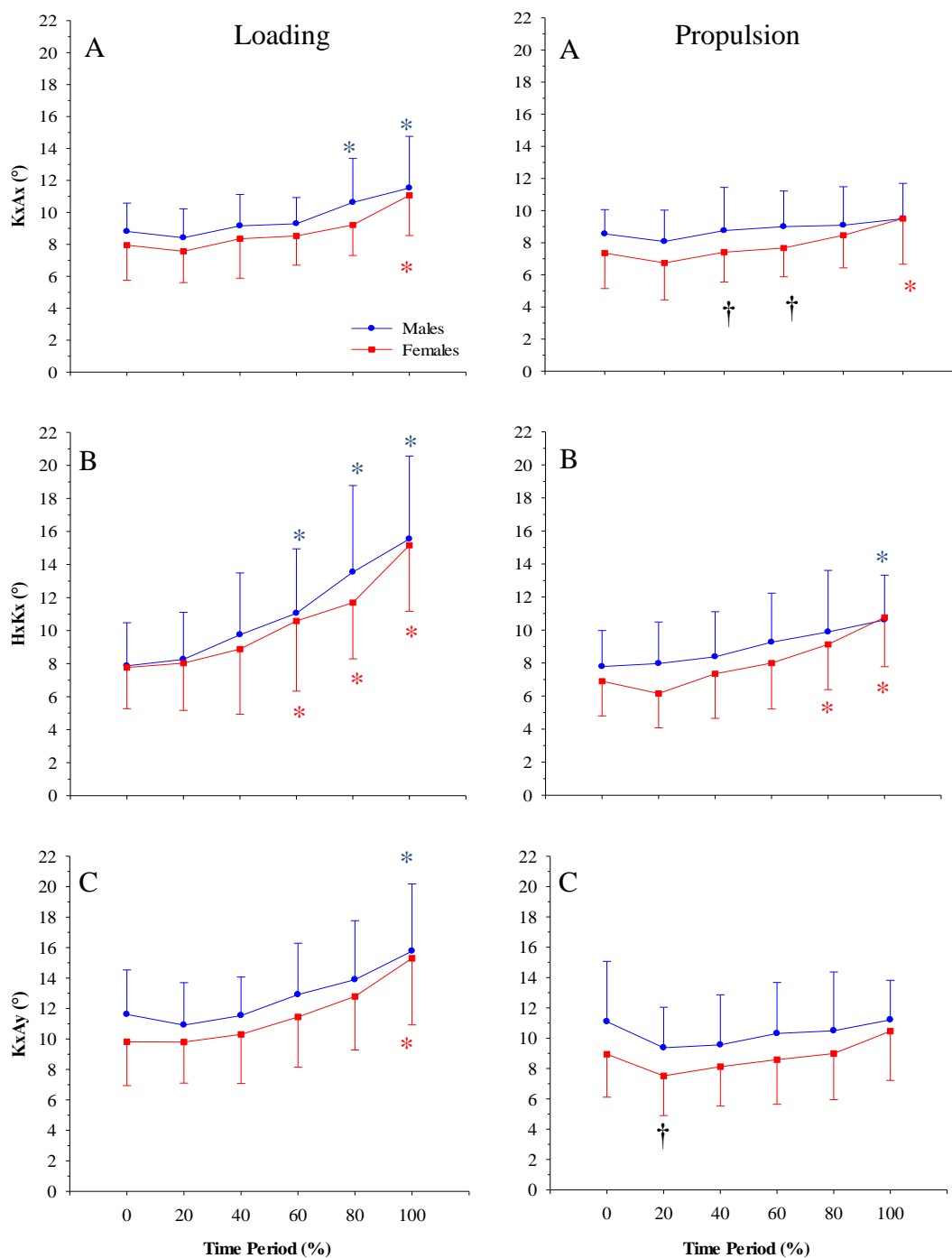


Figure 6.2 Joint coupling variability ($^{\circ}$) (mean (SD)) during loading (left) and propulsion (right) for men (blue) and women (red) at each time period. A) Knee flexion/extension–ankle flexion/extension (KxAx); B) hip flexion/extension–knee flexion/extension (HxKx); and C) knee flexion/extension–ankle abduction/adduction (KxAy). * = $p < 0.05$ compared to 0% time period for men (blue) and women (red). † = $p < 0.05$ between men and women at the specific time period.

6.3.4 Neuromuscular characteristics

6.3.4.1 *Differences between men and women*

Main effects for sex were observed for the total duration of SOL muscle activity (Appendix O.10). Main effects for sex were also observed for MAV of the GM and VL muscles during the feedforward phase (Appendix O.11), the VL and SOL muscles during the background phase (Appendix O.12), the VL and FL muscles during the M2 phase (Appendix O.14), the GM, BF, VL, SOL and FL muscles during the M3 phase (Appendix O.15) and the BF muscle during the propulsion phase (Appendix O.16). Further, there was a significant main effect for sex observed for the SOL:TA co-activation ratio during the M2 phase (Appendix O.17).

Following significant main effects for sex, *post-hoc t*-test results revealed there was a significantly greater increase in MAV of the VL in women compared to men during the feedforward phase at the 40% ($p = 0.016$, $d = 0.85$), 60% ($p = 0.010$, $d = 0.92$), 80% ($p = 0.008$, $d = 1.04$) and 100% ($p = 0.001$, $d = 1.35$) time periods. A significantly greater increase in MAV of the VL during the background phase was displayed by women compared to men at the 20% ($p = 0.008$, $d = 0.89$), 40% ($p = 0.025$, $d = 0.75$), 60% ($p = 0.004$, $d = 0.98$), 80% ($p = 0.020$, $d = 0.77$) and 100% ($p = 0.001$, $d = 1.17$) time periods. There was a significantly greater increase in MAV of the VL during the M2 phase in women compared to men at the 60% ($p = 0.014$, $d = 0.82$), 80% ($p = 0.032$, $d = 0.73$) and 100% ($p < 0.001$, $d = 1.28$) time periods and during the M3 phase at the 80% ($p = 0.013$, $d = 0.83$) time period (Figure 6.3).

There was a significantly greater increase in MAV of the BF in women compared to men during the M3 phase at the 20% ($p = 0.023$, $d = 0.77$), 40% ($p = 0.014$, $d = 0.84$), 80% ($p = 0.007$, $d = 0.91$) and 100% ($p = 0.006$, $d = 0.92$) time periods, and the propulsion phase at the 40% ($p = 0.025$, $d = 0.75$), 60% ($p = 0.046$, $d = 0.67$), 80% ($p = 0.019$, $d = 0.78$) and 100% ($p = 0.028$, $d = 0.72$) time periods (Figure 6.4). A significantly shorter total duration of SOL muscle activity in women compared to men occurred at the 0% ($p < 0.001$, $d = -1.258$), 20% ($p = 0.005$, $d = -0.958$), 40% ($p = 0.011$, $d = -0.864$) and 80% ($p = 0.007$, $d = -0.912$) time periods (Table 6.3). The MAV of the SOL significantly increased in women compared to men during the background phase at the 60% ($p = 0.009$, $d = 0.754$) and 100% ($p = 0.042$, $d = 0.68$) time periods and during the M3 phase at the 40% ($p = 0.020$, $d = 0.77$) time period. Women demonstrated a significantly greater increase in MAV of the GM compared to men during the feedforward phase at the 20% ($p = 0.011$, $d = 0.88$) and 100% ($p = 0.010$, $d = 0.89$) time periods and the MG muscle during the M3 phase at the 100% ($p = 0.031$, $d = 0.73$) time period.

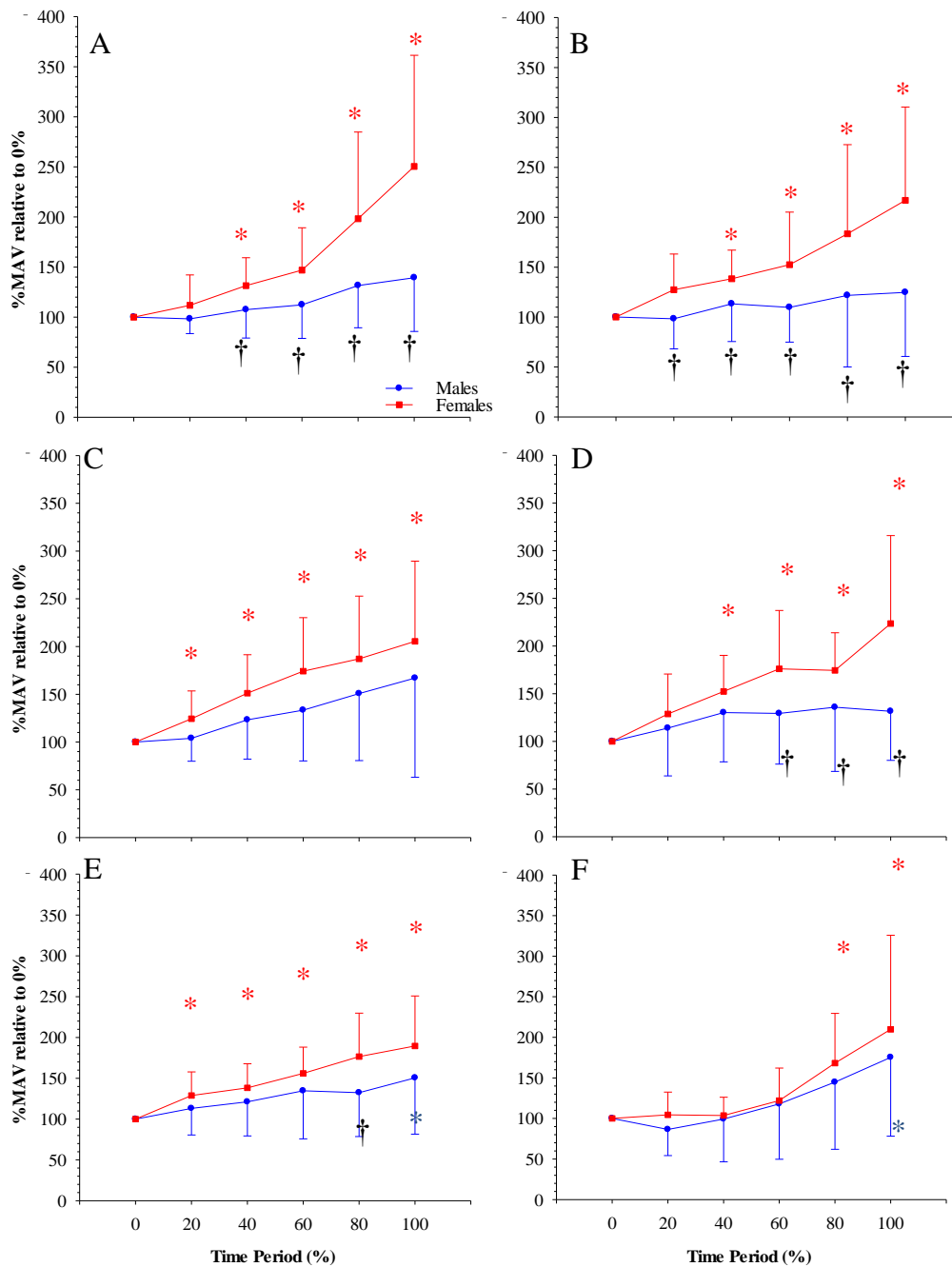


Figure 6.3 Percentage changes in MAV of the VL muscle relative to the 0% time period during the, A) feedforward, B) background, C) M1, D) M2, E) M3 and F) propulsion phase for men (blue) and women (red). The grey line represents baseline (0% time period), * = $p < 0.05$ compared to 0% time period for men (blue) and women (red), † = $p < 0.05$ between men and women at the specific time period.

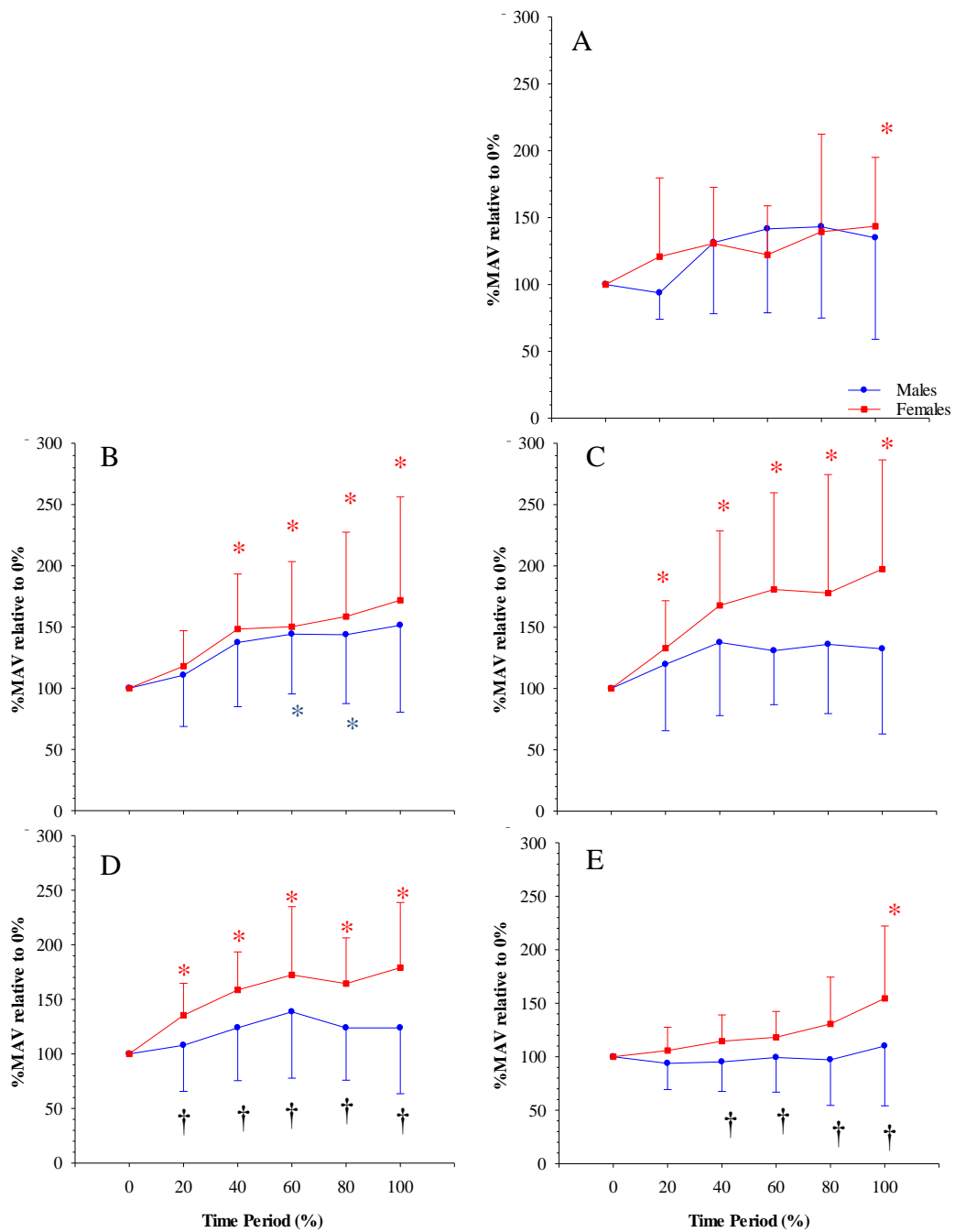


Figure 6.4 Percentage changes in MAV of the BF muscle relative to the 0% time period during the, A) background, B) M1, C) M2, D) M3 and E) propulsion phase for men (blue) and women (red). The grey line represents baseline (0% time period), * = $p < 0.05$ compared to 0% time period for men (blue) and women (red), † = $p < 0.05$ between men and women at the specific time period.

6.3.4.2 *Differences between time periods for men and women*

Main effects for time were observed for the duration of feedforward activity for the GM, MG, SOL and FL muscles and total duration of activity for the VL, MG, SOL and FL muscles (Appendix O.9; Appendix O.10). Time effects were also observed for the MAV of the GM, VL, MG, SOL and FL muscles during the feedforward phase (Appendix O.11), the GM, BF, VL, SOL and FL muscles during the background phase (Appendix O.12), all muscles during the M1 (Appendix O.13), M2 (Appendix O.14) and propulsion phases (Appendix O.16) and the GM, BF, VL, SOL, FL and TA muscles during the M3 phase (Appendix O.15). There was a significant main effect for the SOL:TA co-activation ratio during the background, M1 and M2 phases (Appendix O.17).

Post-hoc one-way ANOVA results revealed only men demonstrated a significant decrease in duration of the feedforward phase of the SOL muscle ($F = 7.622, p < 0.001$) and total duration of the SOL ($F = 7.252, p < 0.001$) and MG ($F = 7.838, p < 0.001$) muscles as time progressed. In contrast, only women demonstrated a significant decrease in the duration of the feedforward phase of the FL ($F = 5.038, p < 0.001$) muscle as time progressed. Common to both sexes, there was a significant decrease in the duration of the feedforward phase of the MG muscle ($F = 20.357, p < 0.001$; $F = 26.109, p < 0.001$) and total duration of the FL muscle ($F = 2.383, p = 0.044$; $F = 4.314, p = 0.002$) as time progressed, respectively (Table 6.3).

Table 6.3 Temporal characteristics (mean \pm SD) for the muscles acting at the hip, knee and ankle joints at all time periods.

		Sex	0%	20%	40%	60%	80%	100%
Duration of feedforward phase (ms)	GM	Men	79.51 \pm 20.64	79.21 \pm 24.00	75.87 \pm 22.40	73.10 \pm 18.49	73.38 \pm 21.87	69.77 \pm 23.13
		Women	80.07 \pm 20.72	76.93 \pm 31.00	77.60 \pm 21.11	78.82 \pm 24.77	82.38 \pm 29.24	64.35 \pm 19.35 *
	VL	Men	45.41 \pm 24.22	51.53 \pm 22.22	52.10 \pm 22.71	52.54 \pm 22.09	52.92 \pm 25.41	54.87 \pm 26.04
		Women	55.61 \pm 32.27	57.73 \pm 24.93	55.37 \pm 25.50	63.06 \pm 26.24	63.49 \pm 24.85	60.72 \pm 21.51
	MG	Men	98.34 \pm 17.48	90.46 \pm 18.21	83.06 \pm 18.22 *	76.73 \pm 15.41 *	71.80 \pm 10.45 *	72.17 \pm 17.54 *
		Women	94.42 \pm 19.75	91.69 \pm 24.92	85.56 \pm 25.47	78.58 \pm 26.10 *	71.91 \pm 24.76 *	72.05 \pm 19.24 *
	SOL	Men	87.38 \pm 23.14	79.35 \pm 21.14	76.63 \pm 24.38	68.15 \pm 21.22 *	71.68 \pm 18.45 *	71.65 \pm 16.63
		Women	70.37 \pm 24.80 ^a	70.91 \pm 24.24 ^a	62.93 \pm 22.43 ^a	62.51 \pm 25.33	60.08 \pm 25.86 ^a	62.01 \pm 22.28
	FL	Men	82.57 \pm 29.71	65.26 \pm 46.71	76.44 \pm 29.77	74.10 \pm 22.59	70.58 \pm 29.38	66.88 \pm 23.28
		Women	79.49 \pm 44.03	70.42 \pm 44.09	67.35 \pm 43.54	60.00 \pm 38.65	63.74 \pm 38.62	51.57 \pm 36.58 *
Total duration of muscle activity (ms)	GM	Men	302.88 \pm 28.59	301.73 \pm 27.45	301.97 \pm 26.05	299.71 \pm 28.85	308.42 \pm 30.18	314.92 \pm 25.21
		Women	301.48 \pm 30.56	313.54 \pm 26.16	311.96 \pm 31.73	316.75 \pm 21.22	330.23 \pm 26.31	313.55 \pm 32.60
	VL	Men	270.18 \pm 36.62	266.34 \pm 34.62	266.41 \pm 28.51	270.19 \pm 35.40	277.78 \pm 40.58	280.30 \pm 30.04
		Women	279.49 \pm 37.68	277.66 \pm 40.03	269.51 \pm 32.42	287.71 \pm 39.50	295.47 \pm 34.34	299.34 \pm 38.03
	MG	Men	344.73 \pm 18.76	334.69 \pm 24.36	327.20 \pm 26.46	320.39 \pm 25.71 *	320.29 \pm 23.96 *	330.71 \pm 23.78
		Women	333.57 \pm 22.85	334.28 \pm 22.50	320.97 \pm 26.36	317.55 \pm 25.54	323.02 \pm 26.30	327.82 \pm 24.21
	SOL	Men	342.05 \pm 26.13	333.78 \pm 24.35	321.94 \pm 29.92 *	315.01 \pm 25.46 *	325.42 \pm 24.39	331.22 \pm 27.75
		Women	310.65 \pm 23.81 ^a	309.32 \pm 26.71 ^a	297.14 \pm 27.47 ^a	302.62 \pm 28.30	303.47 \pm 23.72 ^a	314.90 \pm 30.57
	FL	Men	342.24 \pm 27.79	323.04 \pm 40.59	324.07 \pm 32.02 *	322.50 \pm 30.63	323.78 \pm 31.83	326.64 \pm 24.67
		Women	333.46 \pm 28.47	320.94 \pm 36.52	314.03 \pm 30.18	308.55 \pm 29.99 *	317.66 \pm 29.40	307.78 \pm 33.13
	TA	Men	162.41 \pm 31.80	155.33 \pm 35.61	157.88 \pm 30.63	146.46 \pm 23.80	146.07 \pm 31.95	138.43 \pm 25.30
		Women	185.22 \pm 47.25	164.03 \pm 26.02	172.12 \pm 41.98	152.21 \pm 25.11	160.60 \pm 27.77	163.05 \pm 36.76

* $p < 0.05$ compared to the 0% time period; ^a $p < 0.05$ between men and women at the specific time period

Only women demonstrated a significant increase in the MAV of the VL muscle during the feedforward ($F = 21.998, p < 0.001$), background ($F = 15.932, p < 0.001$), M1 ($F = 19.341, p < 0.001$) and M2 ($F = 23.019, p < 0.001$) phases as time progressed. Both sexes demonstrated a significant increase in MAV of the VL muscle during the M3 ($F = 8.040, p = 0.002$; $F = 22.039, p < 0.001$) and propulsion phases ($F = 9.668, p < 0.001$; $F = 15.612, p < 0.001$), respectively (Figure 6.3).

Women demonstrated a significant increase in the MAV of the BF muscle during the background ($F = 2.918, p = 0.042$), M2 ($F = 12.447, p < 0.001$), M3 ($F = 17.700, p < 0.001$) and propulsion ($F = 8.216, p < 0.001$) phases as time progressed. Both sexes demonstrated a significant increase in the MAV of the BF muscle during the M1 phase ($F = 5.693, p < 0.001$; $F = 8.054, p = 0.001$) as time progressed, respectively (Figure 6.4).

There was a significant increase in the MAV of the FL muscle for women during the background ($F = 8.125, p < 0.001$), M1 ($F = 10.156, p < 0.001$), M2 ($F = 12.354, p < 0.001$) and M3 ($F = 4.316, p = 0.010$) phases as time progressed. In addition, men and women demonstrated a significant increase in the MAV of the FL muscle during the feedforward phase ($F = 5.651, p = 0.004$; $F = 14.409, p < 0.001$) as time progressed, respectively (Figure 6.5).

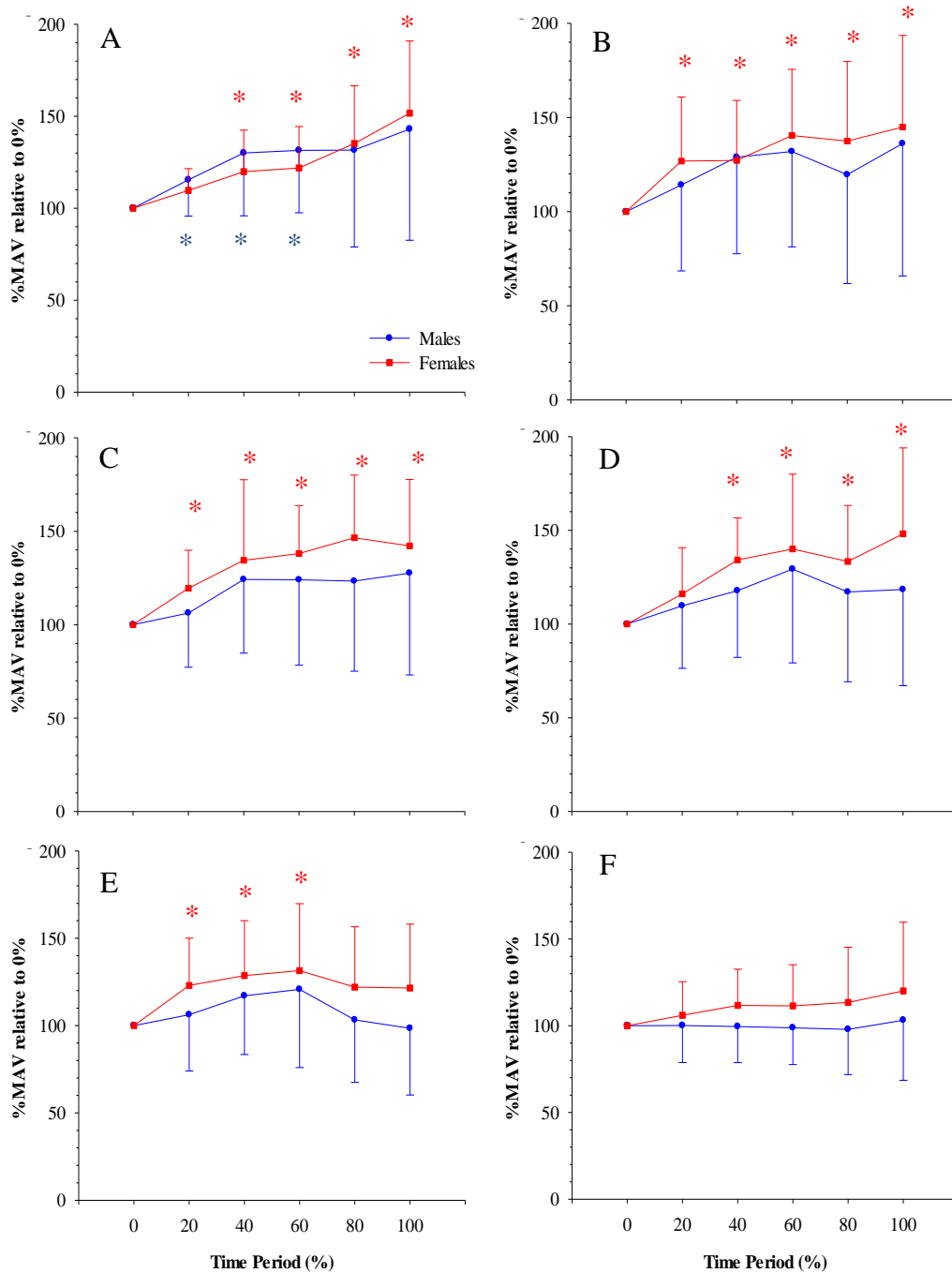


Figure 6.5 Percentage changes in the MAV of the FL muscle relative to the 0% time period during the, A) feedforward, B) background, C) M1, D) M2, E) M3 and F) propulsion phase for men (blue) and women (red). The grey line represents baseline (0% time period), * = $p < 0.05$ compared to 0% time period for men (blue) and women (red).

Only women demonstrated a significant increase in the co-activation ratio of the SOL:TA muscle pair during the background ($F = 19.485, p < 0.001$) and M2 ($F = 6.195, p < 0.001$) phases as time progressed (Figure 6.6). Additionally, both men and women demonstrated a significant increase in the co-activation ratio of the SOL:TA muscle pair during the M1 phase ($F = 3.213, p = 0.045$; $F = 16.117, p < 0.001$), respectively.

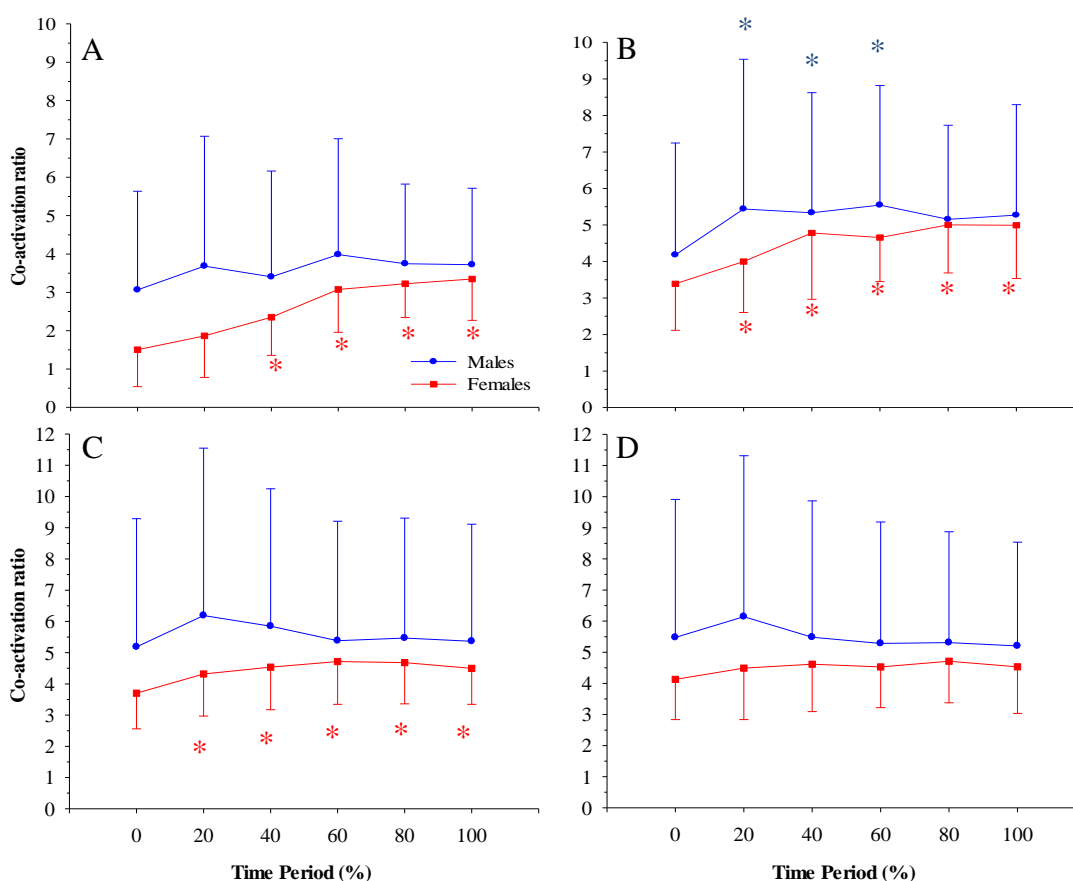


Figure 6.6 Co-activation ratio of SOL:TA during the, A) background, B) M1, C) M2 and D) M3 phase for men (blue) and women (red) at each time period. * = $p < 0.05$ compared to 0% time period for men (blue) and women (red).

Common to both sexes, there was a significant increase in the MAV of the SOL muscle during the feedforward ($F = 13.588, p < 0.001$; $F = 16.411, p < 0.001$), background ($F = 11.775, p < 0.001$; $F = 15.644, p < 0.001$), M1 ($F = 24.548, p < 0.001$; $F = 26.711, p < 0.001$), M2 ($F = 10.630, p < 0.001$; $F = 17.522, p < 0.001$) and M3 ($F = 4.666, p = 0.012$; $F = 14.002, p < 0.001$) phases as time progressed, respectively (Figure 6.7).

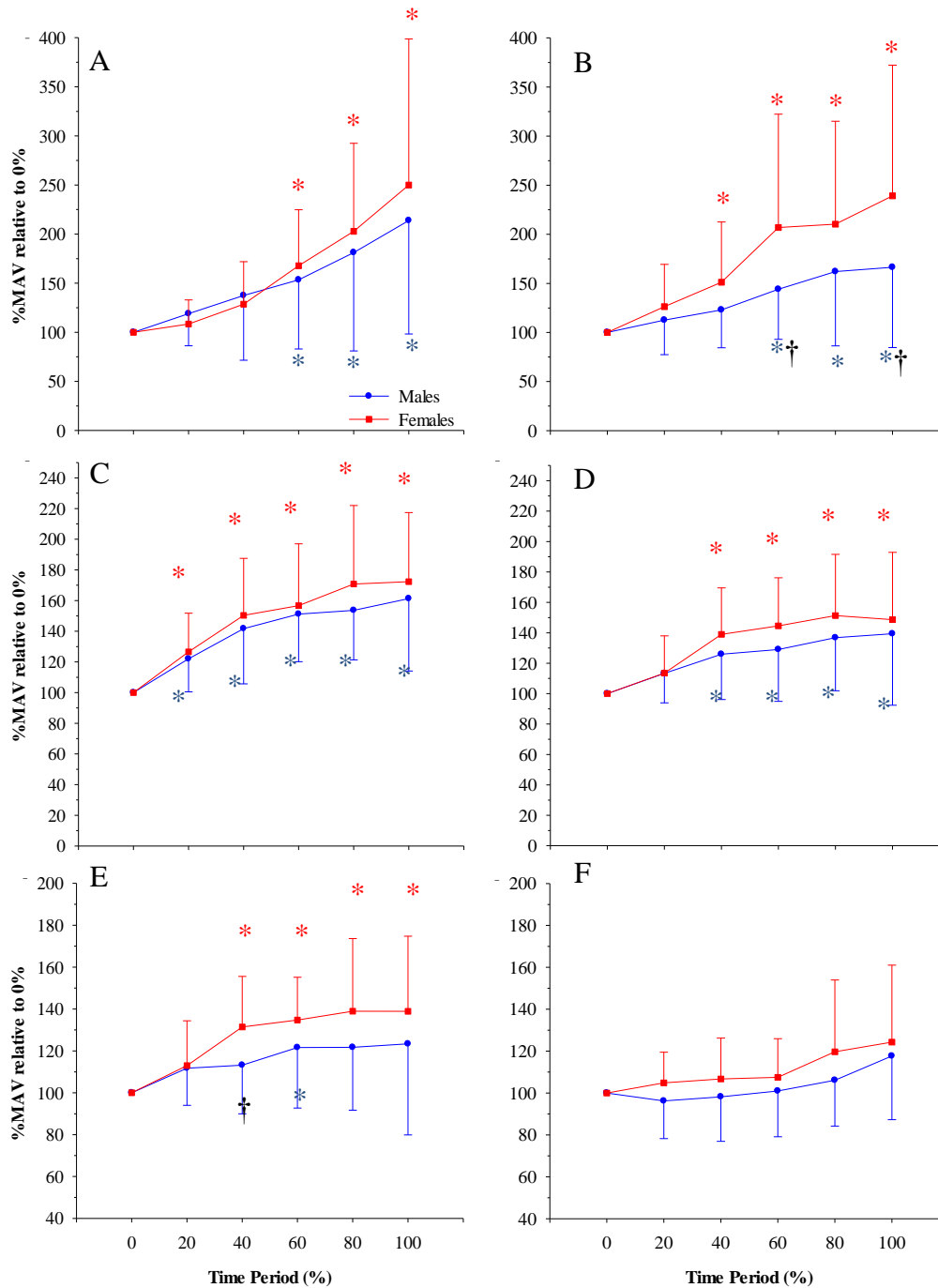


Figure 6.7 Percentage changes in the MAV of the SOL muscle relative to the 0% time period during the, A) feedforward, B) background, C) M1, D) M2, E) M3 and F) propulsion phase for men (blue) and women (red). The grey line represents baseline (0% time period), * = $p < 0.05$ compared to 0% time period for men (blue) and women (red), † = $p < 0.05$ between men and women at the specific time period.

Men and women demonstrated a significant increase in the MAV of the GM muscle during the M2 ($F = 13.066, p < 0.001$; $F = 19.885, p < 0.001$), M3 ($F = 9.973, p < 0.001$; $F = 12.253, p < 0.001$) and propulsion ($F = 11.407, p < 0.001$; $F = 17.488, p < 0.001$) phases as time progressed, respectively. Only women demonstrated a significant increase in the MAV of the GM muscle during the feedforward ($F = 6.190, p = 0.001$), background ($F = 4.887, p = 0.006$) and M1 ($F = 6.759, p < 0.001$) phases as time progressed.

There was a significant increase in the MAV for men and women for the MG muscle during the M1 phase ($F = 3.504, p = 0.024$; $F = 5.315, p = 0.003$) and the TA muscle during the propulsion phase ($F = 10.518, p < 0.001$; $F = 10.128, p < 0.001$) as time progressed. However, only women demonstrated a significant increase in the MAV of the MG muscle during the propulsion phase ($F = 7.103, p = 0.003$) as time progressed.

6.4 Discussion

This study examined the effect of sex on neuromechanical characteristics during a repetitive loading task to exhaustion. The main findings of this study were that although men had a greater amount of movement variability when the single-leg hopping task commenced, as the task progressed women had a greater increase in variability compared to men such that joint coupling variability was similar between the sexes by the end of the task. Further, women had a greater relative dependence on the quadriceps muscle group as demonstrated by a greater increase in VL activity compared to men while the change in MG and SOL activity was similar between the sexes. These findings suggest that during the continued performance of a repetitive loading task, men and women regress to similar patterns of joint coupling variability. However, as fatigue increased, women adopted a greater quadriceps dominant strategy compared to men to allow the maintenance of similar leg mechanics during the exhaustive task, which were no different between the sexes.

Performance parameters remained similar between sexes throughout the single-leg hopping task, with no differences in hopping frequency, hopping height and vertical stiffness. The stereotypical performance was deemed to be sensitive to allow comparison of any detectable differences in kinetics, kinematics and muscle activation rather than being influenced by differences in the technique of hopping

between the sexes. A specific aim was to determine differences in a task that induced fatigue while there was sustained performance, demonstrated by a significant decrement in the maximal height of hopping following the exhaustive trial for both sexes.

In the current study, women had significantly lower levels of joint coupling variability during the propulsion phase compared to men, in the early and mid-stages of the exhaustive trial. These findings were in agreement with previous research (Barrett et al., 2008; Pollard et al., 2005) that demonstrated women to display lower variability than men during a non-fatigued cut and running task. A key distinction from previous research was that the current task was performed to exhaustion, whereas previous research was non-fatiguing (Barrett et al., 2008; Pollard et al., 2005). Although performance output was similar between sexes during the trial, as fatigue progressed women increased their joint coupling variability by a greater magnitude than men. This resulted in the sexes regressing to similar levels of joint coupling variability by the end of the task as fatigue progressed. Previous research has suggested that lower joint coupling variability may lead to a more localised loading of lower limb musculature (Hamill et al., 1999; Heiderscheit et al., 2002). The finding that women had lower joint coupling variability during the first half of the trial may potentially expose women to a greater number of cycles with similar loading to tissues within the kinetic chain compared to men. If low joint coupling variability is considered detrimental to the musculoskeletal system, it is possible women may aim to adapt variability levels to a more 'optimal' range as the task progressed. Therefore, although women may inherently possess lower joint coupling variability than men, this study demonstrated they have the capacity to achieve similar values under particular conditions.

In the current study, differences between the sexes in joint coupling variability were smaller in magnitude compared to previous research by Pollard et al. (2005) that had evaluated a cut task. It is likely both the lower magnitude of difference between sexes and lower absolute values of joint coupling variability in this study, compared to this previous research, may be related to the task being performed. Maintaining a preferred hopping frequency of 2.2 Hz to a target hop height reduces the joint angular displacements and the number of joints involved in generating positive work compared to a cut task (Lamontagne & Kennedy, 2013). These differences between

tasks result in single-leg hopping being a simpler, less complex activity. The lower complexity of the hopping task will most likely result in a smaller influence from any potential differences in skill level which may have affected technique. The current results demonstrated only subtle differences in joint coupling variability are likely to be observed between the sexes during a simple, controlled repetitive task performed to exhaustion. Therefore, differences in joint coupling variability between sexes may be sensitive to the type and complexity of the task being performed.

Importantly, both sexes completed the task for the same duration and maintained a similar and consistent hopping technique, demonstrated by the near identical duration of hopping, spatio-temporal characteristics and joint kinetics and kinematics between men and women during the exhaustive task. A similar decrease in knee joint ROM and increase in knee joint stiffness during loading as the task progressed was observed between sexes. These findings highlight that changes in the movement patterns as fatigue progressed occurred concurrently between sexes at the knee joint. Thus, revealing that individuals completing the same task display the same performance output characteristics utilising the same movement patterns irrelevant of sex.

Evident in the current study, women demonstrated a significantly greater increase in VL muscle activity as the task progressed compared to men, whereas increased SOL activity over the duration of the task was similar between the sexes. These results revealed women were more dependent on the knee musculature during the loading phase to maintain performance output during single-leg hopping compared to men. Sex differences in muscle activity occurred during the centrally modulated feedforward, background and M3 (functional-stretch reflex) phases, and the peripherally modulated M2 (long-latency stretch reflex) phase. Women in this study adapting a greater quadriceps dominant strategy compared to men is in agreement with previous research during drop jumps (Griffin et al., 2006; Nagano et al., 2007; Shultz et al., 2009; Urabe et al., 2005). The current study extends the body of research demonstrating a quadriceps dominant strategy by women to become more pronounced as fatigue progresses. It is speculated these differences are modulated by both central and peripheral control centres.

While previous research has revealed a quadriceps dominant strategy during drop jumps/landings (Nagano et al., 2007; Shultz et al., 2009; Urabe et al., 2005), the current study supported these findings during a primarily ankle modulated hopping task. This suggests women adopt a quadriceps dominant strategy across different types of loading tasks, and as fatigue progresses, and thus in comparison to men, they demonstrated an altered modulation of their movement strategy to achieve a similar output as fatigue increased.

In contrast to the greater increase in VL activity shown by women compared to men, both sexes modulated the primary ankle extensor muscles (MG and SOL) as well as the short-latency stretch reflex (M1 phase) for all measured lower limb muscles similarly. Thus, it appears both men and women modulate spinal-reflex activity similarly during rapid-loading tasks performed to exhaustion. The invariant nature between sexes in triceps surae muscle activity and the short-latency stretch reflex over the duration of the exhaustive task may highlight the primary role of each muscle and its function in achieving the task performance output. Whereas sex differences that occurred in lower limb muscle activity during the feedforward, background, M2, M3 and propulsion phases may highlight that compensatory neuromuscular control strategies are of a primary supra-spinal or central origin.

Despite similar changes in the modulation of the stretch-reflex in the triceps surae muscles, women have previously demonstrated lower active muscle stiffness (Granata, Padua, & Wilson, 2002; Granata, Wilson, & Padua, 2002) and increased tendon compliance (Bryant et al., 2008; Eiling, Bryant, Petersen, Murphy, & Hohmann, 2007). These changes in muscle-tendon function may be attributed to an estrogen induced effect on collagen synthesis (Bryant et al., 2011). As active muscle stiffness and tendon compliance contribute to vertical stiffness (Farley et al., 1991; Farley & Gonzalez, 1996), Bryant et al., (2011) hypothesised that women may modulate neuromuscular control to achieve similar values of lower limb stiffness to men of the same body mass. Therefore, greater VL activation in the current study is hypothesised to be a central or supra-spinal compensatory neuromuscular strategy necessary for women to achieve adequate lower limb stiffness.

6.5 Conclusion

This study highlighted there are both common and unique adaptation strategies between men and women during an exhaustive task when the goal was to maintain performance output. As fatigue progressed there were no differences in total task duration or movement patterns between sexes. In contrast, women had a greater increase in quadriceps activity which paralleled a greater increase in joint coupling variability. These findings highlight a difference between the sexes in how lower limb movement was modulated, but not the resultant movement patterns, during single-leg hopping as fatigue increased.

CHAPTER SEVEN

General discussion

7.1 General discussion

Human locomotion is commonly performed for prolonged durations, exposing the system to a progressive development of fatigue. To allow the continuous performance of repetitive and rhythmical bounding gaits there are alterations to neuromechanical characteristics which are task specific. However, the relationship between changes in neuromechanical characteristics when performance output is maintained remains unknown due to differences in testing protocols and sample populations in previous studies. Understanding how individuals adapt to the development of fatigue during dynamic tasks has the ability to provide insight into the control strategies used to mitigate fatigue during a continuous task. Therefore, the purpose of this thesis was to examine the effect of local muscle fatigue on neuromechanical adaptations and whether these changes were similar between sexes when maintaining performance of a continuous lower limb loading task. To achieve this, a preferred measure of performance output during single-leg hopping was first determined (Chapter 3). Subsequently, the effect of fatigue on joint coupling variability (Chapter 4) and neuromuscular activity (Chapter 5) when maintaining performance during single-leg hopping was assessed. Finally, the effect of sex on neuromechanical characteristics during exhaustive single-leg hopping was determined (Chapter 6).

The first aim of this thesis was to identify a robust and repeatable measure of the vertical displacement of the COM and K_{vert} during continuous single-leg hopping (Chapter 3). These measures served as performance output measures during on-the-spot single-leg hopping. To ensure performance was successfully maintained during the exhaustive trial it was necessary to ensure the method used to measure performance output was suitable. The results of Appendix A determined the segmental analysis method had the greatest validity in determining K_{vert} and vertical displacement of the COM during single-leg hopping. Subsequently, this method was used to measure performance output during exhaustive single-leg hopping (Chapters 4 - 6).

The results presented in Chapter 4 demonstrated an increase in joint coupling variability as muscle fatigue progressed during a dynamic task while performance output was maintained. Theoretically, movement variability provides flexibility to

the neuromechanical system to allow adaptations to potential changes in task demands (Bernstein, 1967; Hamill et al., 2012; Latash, 2012). Further, consistent high level performance across a variety of changing task constraints has been suggested to only be achieved with an optimal level of movement variability (Hamill et al., 2012; Latash, 2000, 2012). The findings of an increase in hip-knee and knee-ankle variability as fatigue progressed (Chapter 4), may be a mechanism to achieve an optimal range of variability in movement based on the demands of the task and energy available to perform the task. Without the ability to provide flexibility within the neuromechanical system, the individual may be less adaptable to intrinsic or extrinsic changes during the task possibly leading to an early cessation of the task, injury or decreased performance. For example, it is plausible that low levels of variability may result in repetitively loading the same musculoskeletal structures over time, greater focal stress and subsequent injury (Bartlett et al., 2007; Hamill et al., 1999; Meardon et al., 2011; Preatoni et al., 2013).

Changes in neuromuscular activity during an exhaustive single-leg hopping trial were investigated and demonstrated an increase in activity of the prime movers including the SOL and VL muscles (Chapter 5). These increases in muscle activity may have been a strategy to overcome greater muscle fatigue such that an increased number of motor units were recruited to maintain ankle and knee function, respectively. The significant increase in SOL muscle activity suggests that the ankle joint remained the predominant site for modulation of single-leg hopping at submaximal and continuous efforts. In addition there was also a significant increase in the activity of the VL muscle as fatigue progressed. However, an increase in VL muscle activity predominately occurred in women and not men, despite no differences in performance output and a parallel increase in knee stiffness (decreased knee ROM) during loading in both sexes. This suggests that compared to men, women may become more reliant on the knee musculature to achieve the same hopping task as fatigue progressed.

The ankle joint is proposed to modulate lower limb mechanical behaviour during hopping when there are changes in extrinsic factors such as cadence (Farley & Morgenroth, 1999), surface stiffness (Farley et al., 1998) and whether performed as a single-leg or double-leg movement. In contrast, the knee joint has been observed to provide a greater contribution to performance during maximal height hopping

(Hobara et al., 2009) and counter-movement jumps (Horita, Komi, Nicol, & Kyrolainen, 2002). Thus, either single or repeated effort landings that result in greater vertical displacement of the COM and ground reaction forces represent activities requiring greater effort and therefore greater contribution of mechanical work about the knee (Hobara et al., 2009; Horita et al., 2002). The current findings suggest the role of knee joint function increased as fatigue progressed when there was potentially greater effort required to maintain the task. In the current study during submaximal efforts to exhaustion, the knee became more rigid, possibly allowing the ankle to optimise its ability to both absorb force and propel the body when hopping. The finding of greater VL muscle activity during the loading phase may have contributed to greater knee stiffness. The concomitant increase in SOL muscle activity allowed for the absorption of force during loading and maintenance of propulsion to the target height throughout the single-leg hopping efforts that were performed to exhaustion. Overall, the current findings demonstrate a synergistic change in VL and SOL muscle activity to modulate fatiguing hopping performance, which were distinctive between sexes.

Typically, movement variability and neuromuscular activity are investigated independent of one another. The muscles of the human system are the effectors responsible for movement of the bony segments, and thus changes in joint coupling variability is likely a by-product of adaptations in muscle activity (Samaan et al., 2015). Chapters 4 and 5 collectively demonstrated inter-dependence between movement variability and neuromuscular activity during an exhaustive dynamic lower limb task when maintaining performance output. Specifically, greater variability in movement between joints may be related to altered muscle activation characteristics within and between muscles. This relationship is supported by the notion of increased variability between segments with the concomitant finding of an increase in lower limb muscle activity as fatigue progressed. In support, Chapter 6 found women demonstrated a greater increase in HxKx and KxAx joint coupling variability and VL muscle activity as fatigue progressed compared to men. It is postulated that increased variability is caused by a greater recruitment of additional muscles to contribute to the maintenance of performance output as fatigue progressed (Srinivasan & Mathiassen, 2012).

A small but statistically significant decrease in vertical stiffness for women but not men may suggest that women were approaching a change in movement pattern defined by the change in leg mechanical characteristics. Considering women had significantly greater increases in VL activity, it is possible that women had to compensate more greatly than men due to increasing fatigue of the SOL muscle by greater relative increases in VL activity. The small decrement in vertical stiffness at the 80% time period in women may have demonstrated the greater level of relative exhaustion of the triceps surae compared to men. Whether there is a causative link between changes in vertical stiffness and injury is not clear, however it is plausible the different compensatory strategies may be influenced by injury and training history.

7.2 Limitations

A common limitation when investigating fatiguing tasks is ensuring participants achieve a certain level of fatigue. It is difficult to accurately quantify the magnitude of fatigue, particularly during dynamic tasks. A delimitation of the current study was participants were instructed to perform the hopping task to volitional exhaustion and fatigue was quantified as a decrease in maximal hopping height following cessation of the exhaustive trial. However, fatigue involves both central and peripheral mechanisms and is a difficult and complex process to quantify.

A method commonly used to quantify fatigue and neuromuscular function during isometric tasks involves the investigation of muscle twitch properties (Gandevia, 2001). This can provide valuable information on changes in levels of central or peripheral fatigue during a continuous task. Integrating these methods into the current study would provide a clearer indication to the contributions to task cessation and further our understanding of the specific neuromuscular processes during fatiguing exercise. However, this method is seldom used during dynamic tasks that cannot be performed within a stationary ergometer. Alternatively, an isometric handgrip strength test completed before and after the fatiguing protocol could have broadly assessed central fatigue.

Hopping mimics bounding gait patterns that utilise the SSC muscle action, such as running and walking by utilising similar muscle contraction types and limb loading behaviours. However, caution should be taken from drawing inappropriate

conclusions due to the oversimplifications of the hopping model (Lamontagne & Kennedy, 2013). For example, hopping has demonstrated differences with other bounding gait patterns such as smaller joint excursions, less contribution from other segments such as the trunk, head and arms and differences in joint moments. Thus, findings during single-leg hopping may not reflect the possible changes in neuromechanical characteristics during running, walking and repeated landing from jumps.

Performing trials within a controlled laboratory environment eliminated many potential environmental influences. This improves the ability to isolate adaptations in neuromuscular characteristics due to changes in intrinsic factors. However, it is not possible to determine the interaction between intrinsic and extrinsic factors from the current results.

There may be potential differences in training and loading histories between sexes as men and women were not matched for the amount or type of activity they had performed prior to testing. A delimitation of the current study was all participants were required to be recreationally active and participate in regular exercise that involved running for at least one to four hours per week. Although this does not predicate that men and women were matched for the amount of current and previous physical activity, there were no differences in the total duration of the task or performance output between sexes.

Participants were required to be injury free during the six months prior to testing. However, it is possible injuries that occurred prior to this time frame may influence the changes in neuromuscular characteristics if he/she had habituated to a change in movement strategy due to the injury. Participants were recreationally active and required to be regularly involved in physical activity, specifically exercises that involved loading of the lower limb. This was chosen to encompass a wide variety of activities which people may have participated in. The results of the current study cannot be extrapolated to elite or sedentary populations, or populations of a different age such as children, middle-aged, or older individuals.

7.3 Future directions

The current program of research provided a basis for understanding the time course of changes in movement variability and control during continuous repeated lower limb loading tasks in healthy individuals. Due to the context specific nature of these measures it is important to assess these adaptation strategies during more complex locomotor tasks, such as walking and running. Further, it remains unclear whether these changes would be consistent when extrinsic environmental factors are present, such as changes in surface. Future research is necessary to provide better insight into adaptation strategies during real world settings where intrinsic and extrinsic factors are prevalent.

The current single-leg hopping task was only completed on the dominant limb and was relatively short in duration compared to other submaximal locomotion tasks that are performed to exhaustion. Future research may aim to investigate whether these changes are consistent between the dominant and non-dominant legs and over a task of a longer duration, or one completed intermittently. From a methodological perspective, an intermittent fatigue protocol would allow a longer temporal profiling of neuromuscular function. Further, the investigation of muscle twitch properties may be interwoven during the task to provide insightful information on changes in central and peripheral fatigue and the contributions of these factors to task cessation.

It is important to determine whether increases in joint coupling variability contributed to the maintenance of performance output. Additionally, determine whether limiting movement variability may influence the time to exhaustion when maintaining performance. Future research is needed to determine if increases in joint coupling variability is beneficial for performance and/or injury risk during exhaustive tasks. This could be achieved by completing controlled studies whereby participants are to either maintain performance or not maintain performance as fatigue progresses. Alternatively, restrictions in movement variability could be imposed. These results may provide further impetus for future training studies to be developed to manipulate joint coupling variability for performance.

An improved understanding between variables of movement variability and traditional biomechanical measures such as mean joint angles and GRF is a necessary step to improve the integration of these methods. Understanding the

relationship between different measurement techniques will allow a subsequent better acceptance of measures of variability within research and practice and improve the interpretation of results related to these techniques. The merging of biomechanics and motor control has the potential to provide a wealth of information to improve the understanding of human movement (Latash, 2016). In particular, how these mechanisms are regulated to adapt to a variety of changing task constraints. Ideally, this will allow a better transfer of information between the researchers and the coaches to improve individual performance and/or reduce injury risk.

Limited research identifying the cause and effect relationship between movement variability and injury risk remains. This can be improved by the completion of rigorous scientific studies of a prospective nature, similar to Maudler et al. (2013). However, this is a challenging task due to many limitations and confounding factors associated with these study designs. Alternatives such as the use of saline injections to induce pain may offer insight into the cause and effect relationship between pain, movement variability and injury. As pain is a function of injury, these study designs can improve the understanding into the role of movement variability within an injury context. In addition, providing information on whether pain is an influencing factor. Importantly, these findings will add to the growing literature to better define what may be considered too high, low or optimal levels of variability during movement, an important factor required in future research.

In addition to injury risk, monitoring recovery from injury can provide a greater context to these biomechanical and motor control measures. Previous research has demonstrated subtle differences retrospectively between healthy, current low back pain and recovering from low back pain patients (Seay et al., 2011). However, there remains no research to date monitoring changes in movement variability prospectively such as during a hopping task, prior to and following injury. This could provide another method for researchers and coaches to help in the prospective monitoring of injury prevention and return to play.

7.4 Conclusion

This thesis demonstrated a temporal relationship between neuromuscular adaptations and changes in joint coupling variability as fatigue progressed during a continuous lower limb loading task when performance was maintained. Men and women had

similar changes in neuromechanical characteristics however, women had a greater increase in joint coupling variability and quadriceps activity as the task progressed. These adaptations were thought to be a strategy to overcome the effects of local muscle fatigue and preserve performance output in healthy participants.

Overall, the results of this thesis suggest that changes in neuromechanical characteristics during stereotypical movements are not detrimental to performance and sexes display similar changes during repetitive, submaximal efforts to exhaustion. Women do demonstrate greater quadriceps activity than men, however, are equally resilient to increasing levels of muscle fatigue. These findings may inform coaches or trainers as to the benefits of movement variability during continuous repeated loading tasks for skill execution and injury prevention. Future research is necessary to contextualise these relationships during functional movements within a sporting and rehabilitation setting.

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APPENDICES

**Appendix A: Mudie et al., (2016). Journal article published in the
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A Comparison of Vertical Stiffness Values Calculated from Different Measures of Center of Mass Displacement in Single-Leg Hopping

Kurt L. Mudie,¹ Amitabh Gupta,¹ Simon Green,¹ Hiroaki Hobara,² and Peter J. Clothier¹

¹Western Sydney University; ²National Institute of Advanced Industrial Science and Technology

This study assessed the agreement between K_{vert} calculated from 4 different methods of estimating vertical displacement of the center of mass (COM) during single-leg hopping. Healthy participants ($N = 38$) completed a 10-s single-leg hopping effort on a force plate, with 3D motion of the lower limb, pelvis, and trunk captured. Derived variables were calculated for a total of 753 hop cycles using 4 methods, including: double integration of the vertical ground reaction force, law of falling bodies, a marker cluster on the sacrum, and a segmental analysis method. Bland-Altman plots demonstrated that K_{vert} calculated using segmental analysis and double integration methods have a relatively small bias ($0.93 \text{ kN}\cdot\text{m}^{-1}$) and 95% limits of agreement (-1.89 to $3.75 \text{ kN}\cdot\text{m}^{-1}$). In contrast, a greater bias was revealed between sacral marker cluster and segmental analysis ($-2.32 \text{ kN}\cdot\text{m}^{-1}$), sacral marker cluster and double integration ($-3.25 \text{ kN}\cdot\text{m}^{-1}$), and the law of falling bodies compared with all methods (17.26 – $20.52 \text{ kN}\cdot\text{m}^{-1}$). These findings suggest the segmental analysis and double integration methods can be used interchangeably for the calculation of K_{vert} during single-leg hopping. The authors propose the segmental analysis method to be considered the gold standard for the calculation of K_{vert} during single-leg, on-the-spot hopping.

Keywords: agreement, lower limb, spring-mass model, kinetic, kinematics

Leg stiffness is often determined during human locomotion when describing the motion of the mass during landing tasks or bouncing gaits.^{1,2} An increase in leg stiffness has been shown with a shorter contact time at a given hopping frequency,^{3,4} or at a greater hopping frequency⁵ and greater running velocity.⁵ Increasing leg stiffness may be a strategy which improves the utilization of stored elastic energy within the musculoskeletal system in the lower extremity.⁶ However, it is suggested that increased or decreased leg stiffness may also lead to a greater risk of injury due to a shift in the stress placed on the bony and soft tissues.⁷ Therefore, an accurate and consistent calculation of leg stiffness is required to allow a determination of a clinically significant change within or between subjects.

Leg and vertical stiffness (K_{vert}) are used synonymously during tasks where there is no horizontal translation, calculated as the quotient of the change in ground reaction force and displacement of the center of mass (COM).⁸ Although ground reaction force can be directly measured from a force platform, it is problematic to accurately measure the vertical displacement of the COM. Consequently, a number of methods have been developed to estimate the displacement of the COM to calculate K_{vert} .^{7,9–13} The most common method used to calculate vertical COM displacement during stationary tasks involves double integration of the vertical acceleration measured from a force platform.^{7,10–12,14–20} Less common methods using a segmental analysis and sacral marker have also been used to estimate COM displacement, but have been applied more often

in tasks requiring horizontal translation^{13,21–24} rather than vertical displacement.¹² A simplified method using flight and contact times to calculate vertical COM displacement has previously been proposed,^{9,25} although not validated against multiple criterion measures.

During double-leg hopping at 2 Hz and 2.4 Hz, Ranavolo et al.¹² determined that motion capture of the sacral marker and the reconstructed pelvis methods produced significantly higher measures of vertical displacement of the COM compared with the segmental analysis and double integration methods. However, the authors¹² did not explore the influence of these differences when computing K_{vert} . Further, Hébert-Losier and Eriksson¹⁰ determined that changing the integration constants used in the double integration method affected the measure of K_{vert} in double-leg hopping by changing the calculated initial velocity and displacement values. Lower K_{vert} values were also calculated when modeling the change in vertical ground reaction force (vGRF) as a sine-wave compared with the natural frequency of oscillation method and double integration method during double-leg hopping.¹¹ However, there is a significant difference in K_{vert} between single- and double-leg hopping tasks.^{10,26} The differences between computational methods for the calculation of K_{vert} may therefore be more or less pronounced in single-leg compared with double-leg hopping. Thus, as most athletic activities are performed unilaterally, differences in K_{vert} based on computations should be investigated in a single-leg loading task.

It remains unknown whether vertical displacement of the COM derived from different methods provides similar calculations of K_{vert} during single-leg hopping. The purpose of this study was to test the agreement between estimates of K_{vert} calculated using 4 different methods derived from either kinetic or kinematic-kinetic measures of the vertical displacement of the COM during single-leg hopping. It was hypothesized that significant differences in the calculation of K_{vert} and vertical displacement of the COM would be observed when using different methods.

Mudie, Gupta, Green, and Clothier are with the School of Science and Health, Western Sydney University, Sydney, NSW, Australia. Hobara is with the Human Informatics Research Institute, National Institute of Advanced Industrial Science and Technology, Tokyo, Japan. Address author correspondence to Kurt L. Mudie at kurt.mudie@wsu.edu.au.

Methods

Participants

Thirty-eight recreationally active males ($n = 20$) and females ($n = 18$) (mean \pm standard deviation [SD] 23.2 ± 2.2 years of age; 1.73 ± 0.06 m in height; 74.0 ± 10.4 kg in mass) volunteered to participate in this study. All participants were healthy and participated in exercise for 1–4 hours per week. Participants had no past or current history of lower limb pathology, injury, pain, lower limb fracture, or surgery within the 6 months before testing. Ethical approval (H1074) was granted by The University of Western Sydney Human Research Ethics Committee and all participants provided written informed consent before testing.

Instrumentation

Kinetic and kinematic data were collected synchronously at a frequency of 1500 Hz and 150 Hz, respectively (First Principles, Version 1.2.4) during single-leg, on-the-spot hopping. Kinetic data were sampled from an AMTI multicomponent 600×400 mm force plate (model BP400600-1000, Watertown, MA) and kinematic data sampled using an Optotrak Certus system (Northern Digital Inc., Waterloo, Canada). All data were processed using Visual 3D (C-Motion, Version 4, Germantown, MD). Each participant was instructed to land on the audible tone of a metronome, set to the preferred hopping frequency for humans at 2.2 Hz (132 beats/min).^{3,18}

Participant Preparation and Data Collection

Following measurement of each participant's height and body mass, a warm-up and familiarization period was completed. This allowed each participant to practice hopping on their self-selected dominant leg, defined as their preferred kicking leg,^{15,27} until he/she could complete the task successfully. All trials were performed barefoot and each participant was instructed to keep their hands on their hips, hop on the ball of their foot, land on the beat of the metronome, minimize contact time, and keep the knee of their nonhopping leg flexed at 90°.

A 7-segment model was used to model the trunk,²⁸ pelvis,²⁹ thigh, shank,³⁰ hindfoot, forefoot,³¹ and hallux³² of the hopping leg. Twenty-one active markers were adhered over the segments, including a single 4-marker cluster (sacrum), three 3-marker clusters (T8 vertebra, distal lateral thigh, and shank), and 8 individual active markers (foot).^{33,34} Fifteen calibration markers were digitized to identify the C7 and T8 spinous processes, suprasternal notch, xiphoid process, left and right acromioclavicular joints, left and right anterior superior iliac spines, left and right posterior superior iliac spines, greater trochanter, medial and lateral femoral epicondyles, and medial and lateral malleoli of the dominant leg. Eight individual active markers (7-mm diameter) (Optotrak Certus, Northern Digital Inc., Waterloo, Canada) were adhered to the posterior, medial and lateral calcaneus, base and head of the first and fifth metatarsals, and the medial proximal phalanx of the great toe. Each participant then completed a single trial of on-the-spot, single-leg hopping for 10 seconds at 2.2 Hz.

Data Processing

All recorded trials were exported from First Principles software as C3D files for processing in Visual 3D. Kinematic marker recordings were inspected to determine missing frames and any missing data interpolated using spline interpolation for up to a maximum

gap of 10 frames. A third-order polynomial was used to fill any gap(s) and the 3 frames before and after the missing frame(s) used to calculate the coefficient of the polynomial equation. Recorded marker positions were then bidirectionally filtered using a fourth-order zero-lag Butterworth filter with an 8 Hz cut-off.^{17,35} Kinetic data were bidirectionally filtered with a fourth-order low-pass Butterworth filter with a 50 Hz cut-off.^{25,36}

Data Analysis

A single hop cycle was defined as a complete flight phase and the subsequent contact phase. The contact phase included the loading (initial contact to peak vGRF) and propulsive (peak vGRF to toe-off) phase. Temporal events including initial contact and toe-off were determined for each hop cycle from the vGRF trace as the first and last point ≥ 10 N,³⁷ respectively. Peak vGRF was defined as the maximal vGRF value during the contact phase.

The first 20 hop cycles were extracted from the trial data for each participant ($n = 760$). Seven hop cycles from 5 separate participants were incomplete or contained missing kinematic data and were removed from the analysis. A final total of 753 complete hop cycles were analyzed from all participants. Dependent variables calculated for all hop cycles included K_{net} and vertical displacement of the COM during flight (ΔCOM_f) and loading phases (ΔCOM_l). Each dependent variable was calculated using the sacral marker cluster, segmental analysis, double integration, and law of falling bodies methods.

Sacral Marker Cluster Method. The vertical displacement of the sacral marker cluster was used to estimate ΔCOM_f and ΔCOM_l .^{12,13}

Segment Analysis Method. The vertical displacement of the COM was calculated as the weighted average of the 7 modeled upper and lower limb segments to estimate ΔCOM_f and ΔCOM_l .^{12,13} The instantaneous position of the COM was estimated using the location and masses of each of the 7 modeled segments (Equation 1):³⁸

$$\text{COM}(t) = \frac{1}{M} \sum_{i=1}^n m_i y_i(t), \quad (1)$$

where n was the total number of segments, including the trunk, pelvis, thigh, shank, and foot (inclusive of the hindfoot, forefoot, and hallux), m_i was the mass of a segment, $y_i(t)$ was the instantaneous location of the COM of a segment, and M was the sum of the masses of the segments. The segment masses and inertial properties were derived from Dempster³⁹ and Hanavan and Ernest,⁴⁰ respectively.

Double Integration Method. Vertical displacement of the COM was estimated by integrating the vertical acceleration of the COM twice to calculate values for ΔCOM_f and ΔCOM_l .^{18,38} Vertical acceleration of the COM was first obtained from the vGRF curve at each frame, $1 \leq i \leq n$ as (Equation 2):

$$a_i = \frac{1}{m} f_i - g, \quad (2)$$

where i was each frame that was less than or equal to n , the total number of frames, m was body mass, and g was the acceleration due to gravity ($9.81 \text{ m}\cdot\text{s}^{-2}$).

Vertical velocity was then obtained by integrating the vertical acceleration at each frame as (Equation 3a and b):

$$v_i = v_0 + a_i \cdot \Delta t, \quad (3a)$$

$$v_{i+1} = v_i + a_{i+1} \cdot \Delta t \text{ (for } 1 \leq i \leq n), \quad (3b)$$

where Δt is the change in time between frames (1/1500 s) and the first integration constant v_0 was the velocity of the COM at initial contact, calculated as (Equation 4):

$$v_0 = -\frac{1}{2} \cdot g \cdot t_f, \quad (4)$$

where t_f was the duration of the flight phase.

Vertical displacement was obtained by integrating the vertical velocity at each frame as (Equation 5):

$$s_{i+1} = s_i + v_{i+1} \cdot \Delta t, \quad (5)$$

where the second integration constant (vertical displacement of the COM at initial contact) was considered zero.

Law of Falling Bodies Method. The vertical displacement of the COM was estimated by the law of falling bodies^{4,23} that assumes a parabolic displacement time curve during the flight and contact phase and there was a change in direction of the COM at midflight and peak vGRF when $v = 0 \text{ m}\cdot\text{s}^{-1}$.

The vertical displacement of the COM during the flight phase was calculated by (Equation 6):

$$\Delta\text{COM}_f = \frac{1}{2} \cdot g \cdot \left(\frac{t_f}{2}\right)^2. \quad (6)$$

The velocity of the COM at initial contact was determined by (Equation 7):

$$v_0 = \sqrt{2 \cdot g \cdot \Delta\text{COM}_f}. \quad (7)$$

The vertical displacement of the COM during the loading phase was then calculated by (Equation 8):

$$\Delta\text{COM}_l = \frac{v_0 + v_f}{2} \cdot t_l, \quad (8)$$

where v_f was the velocity of the COM at peak vGRF ($0 \text{ m}\cdot\text{s}^{-1}$ due to a change in direction assumed at peak vGRF) and t_l was the duration of the loading phase.

Vertical stiffness was then calculated for each method as the quotient of force and displacement (Equation 9):⁷

$$K_{\text{vert}} = \frac{\text{peak vGRF}}{\Delta\text{COM}_l}. \quad (9)$$

Statistical Analyses

To quantify the agreement between methods, Bland-Altman plots were created for each method pair for all 3 dependent variables.⁴¹ Specifically, the difference between the methods was plotted against

their mean for each hop cycle (SigmaPlot, Version 12.5). The bias was determined as the mean difference across all hop cycles and the 95% limits of agreement (LoA) calculated as the mean difference (bias) $\pm 1.96 \times \text{SD}$ of the difference.⁴² To quantify trends of the bias over a range of values, the slope of a regression line was calculated for each Bland-Altman plot. The coefficient of variation for each method was calculated as the SD normalized to the mean and converted to a percentage (%) to determine the variance by each method.

Results

Mean, minimum, and maximum values over all hop cycles were calculated for hopping frequency and peak vGRF (Table 1). Mean K_{vert} was greatest using the law of falling bodies method, followed by double integration, segmental analysis, then sacral marker cluster methods (Table 2). For the calculation of K_{vert} across individual hop cycles, Bland-Altman plots showed that the double integration and segmental analysis methods had the greatest agreement with a relatively small bias ($0.93 \text{ kN}\cdot\text{m}^{-1}$) and 95% LoA (-1.89 to $3.75 \text{ kN}\cdot\text{m}^{-1}$) (Figure 1). Further, the segmental analysis and double integration methods had the lowest coefficient of variation (Table 3) and slope of the regression line. In contrast, a greater bias, 95% LoA, coefficient of variation (Table 3), and slope of the regression line were calculated between the sacral marker cluster and the double integration ($-3.25 \text{ kN}\cdot\text{m}^{-1}$, -6.62 to $0.11 \text{ kN}\cdot\text{m}^{-1}$) and segmental analysis methods ($-2.32 \text{ kN}\cdot\text{m}^{-1}$, -4.40 to $-0.25 \text{ kN}\cdot\text{m}^{-1}$) (Figure 1). The law of falling bodies method revealed a relatively large bias and slope of the regression line when compared with the segmental analysis ($18.19 \text{ kN}\cdot\text{m}^{-1}$), double integration ($17.26 \text{ kN}\cdot\text{m}^{-1}$), and sacral marker cluster ($20.52 \text{ kN}\cdot\text{m}^{-1}$) methods (Figure 1).

Mean ΔCOM_l was lowest using the law of falling bodies method, followed by the double integration, segmental analysis, and then sacral marker cluster method (Table 2). For the calculation of ΔCOM_l across individual hop cycles, the double integration and segmental analysis methods demonstrated the greatest agreement (bias and 95% LoA) (-0.48 cm and -1.96 to 1.00 cm) (Figure 2).

Table 1 Hop frequency (mean [SD], minimum and maximum) and peak vertical ground reaction force (vGRF) over 753 hop cycles

	Mean (SD)	Range
Hop frequency (Hz)	2.22 (0.09)	2.00–2.83
Peak vGRF (kN)	1.77 (0.34)	1.11–2.62

Table 2 Vertical stiffness (K_{vert}) (mean [SD]) and vertical displacement of the center of mass during loading (ΔCOM_l) and flight (ΔCOM_f) phases calculated from the sacral marker cluster, segmental analysis, double integration, and law of falling bodies methods

Method	K_{vert} ($\text{kN}\cdot\text{m}^{-1}$)	ΔCOM_l (cm)	ΔCOM_f (cm)
Sacral marker cluster	15.33 (3.27)	11.62 (1.13)	2.47 (1.15)
Segmental analysis	17.65 (3.58)	10.07 (0.96)	2.01 (0.98)
Double integration	18.58 (3.84)	9.59 (1.07)	2.52 (0.99)
Law of falling bodies	35.85 (7.66)	5.01 (0.76)	2.54 (0.99)

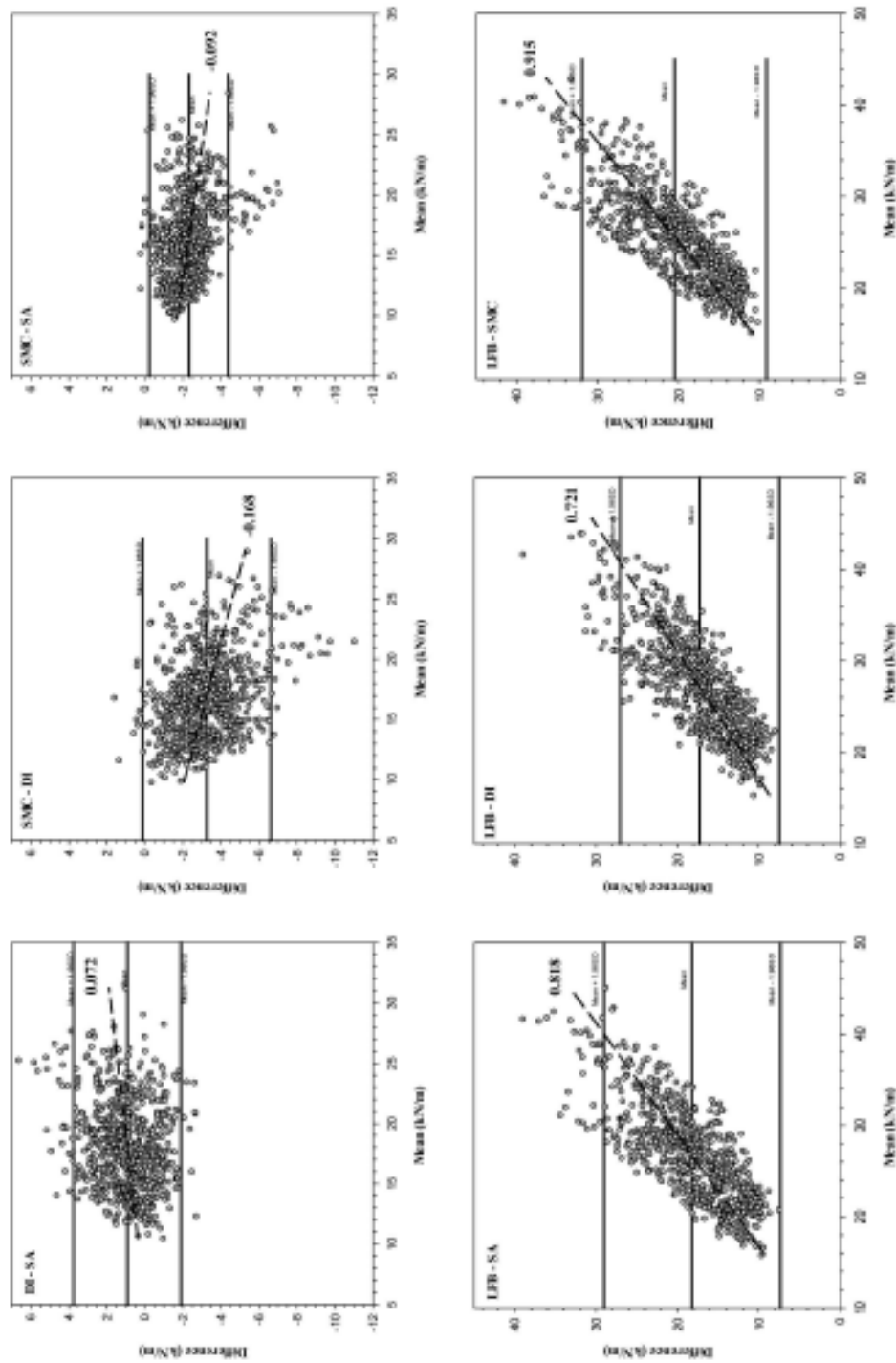


Figure 1 — Bland-Altman plots for the comparison of K_{w0} between the sacral nodal cluster (SMC), segmental analysis (SA), double integration (DI), and low of falling bodies (LFB) methods. The difference (bias) is represented by the inner horizontal solid line, 95% limits of agreement by the outer solid lines, regression line by the dashed line, and the magnitude of the regression slope labeled to the right. The double integration–segmental analysis method pair demonstrates the smallest bias and trend as stiffness increased.

Table 3 Coefficient of variation (%) of the sacral marker cluster, segmental analysis, double integration, and law of falling bodies methods for each dependent variable

Coefficient of Variation (%)	Sacral Marker Cluster	Segmental Analysis	Double Integration	Law of Falling Bodies
Vertical stiffness	21.33	20.27	20.64	21.38
Loading height	9.70	9.54	11.21	15.17
Flight height	46.45	49.10	39.17	39.01

Further, the segmental analysis and double integration methods had the lowest coefficient of variation (Table 3) and slope of the regression line. Bland-Altman plots revealed a relatively large bias but similar 95% LoA between the sacral marker cluster and the segmental analysis (1.55 cm and 0.26–2.84 cm) and double integration methods (2.03 cm and 0.05–4.02 cm). The law of falling bodies method revealed a relatively large bias, coefficient of variation (Table 3), and slope of the regression line when compared with the segmental analysis (–5.06 cm), double integration (–4.58 cm), and sacral marker cluster (–6.61 cm) methods (Figure 2).

Mean $ACOM_L$ was greatest using the law of falling bodies method, followed by the double integration, sacral marker cluster, and then segmental analysis method (Table 2). For the calculation of $ACOM_L$ across individual hop cycles, Bland-Altman plots showed the greatest agreement (bias and 95% LoA) between the law of falling bodies and double integration method pair (0.02 cm and –0.01 to 0.03 cm) (Figure 3). Further, the law of falling bodies and double integration methods had the lowest coefficient of variation (Table 3) and no regression slope. A small bias and 95% LoA were also revealed between the sacral marker cluster and double integration method pair (–0.05 cm and –1.13 to 1.02 cm) and the sacral marker cluster and law of falling bodies method pair (0.08 cm and –1.00 to 1.16 cm). The greatest bias was between the segmental analysis and the sacral marker cluster (0.46 cm), law of falling bodies (0.54 cm), and double integration (0.52 cm) method pairs (Figure 3). There was a relatively large regression slope between the sacral marker cluster and all methods and a small regression slope between the segmental analysis and double integration method.

Discussion

The purpose of this study was to assess the agreement between vertical stiffness calculated from 2 kinetic and 2 kinematic-kinetic methods of estimating the vertical displacement of the COM during single-leg hopping. The main finding of this investigation was that K_{vert} calculated using the segmental analysis and double integration methods were very similar, with the lowest coefficient of variation during a single-leg loading task. The authors propose the segmental analysis method can be considered to represent the gold standard for the calculation of K_{vert} and vertical displacement of the COM during single-leg, on-the-spot hopping. Further, the current results support the use of either the segmental analysis or double integration methods to calculate K_{vert} during single-leg hopping.

When comparing the segmental analysis and double integration methods for the calculation of K_{vert} , Bland-Altman plots revealed only a small bias (< 1 kN·m⁻¹) over a range of stiffness values from 10.58 to 31.12 kN·m⁻¹. Specifically, Bland-Altman plots revealed small 95% LoA and a small proportional systematic error as mean K_{vert} increased between the segmental analysis and double

integration methods. Further, mean K_{vert} from the segmental analysis and double integration methods were most comparable to previous research,^{18,20,26} with lower variance than the sacral marker cluster and law of falling bodies methods.

Assumptions of the segmental analysis method are limited to the segment properties and complexity of the model (number of segments modeled). Modeled segments are assumed to be rigid with dimensions calculated from anthropometric measurements of cadaver specimens.^{12,13,22} Since the calculation of the COM position is based on the weighted average and position of the modeled segments, any movement of nonmodeled segments that would theoretically change COM position cannot be detected. For the current study, since only 1 leg and the trunk were modeled, there was a shift in the position of the COM toward the ipsilateral side. Further, any movement of the arms or contralateral nonweightbearing leg would not be detected. However, the testing protocol aimed to nullify the potential error by minimizing movement of the upper limbs and contralateral nonweightbearing leg. Future researchers investigating single-leg loading tasks may choose to only model the side of interest for ease of testing and to save time during the experimental set-up, eliminating unnecessary data.

The assumptions of the double integration method relate to the calculations which may smooth the displacement signal, potentially explaining the lower COM displacement values compared with the segmental analysis method.¹² In addition, determination of the velocity integration constant assumes symmetry of the hop cycle. Due to the asymmetry of the hop cycle and potential differences within and between subjects, these assumptions are incorrect and will impact the calculated K_{vert} . As the value of the initial velocity integration constant has been shown to produce approximately 14% variance of the calculated K_{vert} values during double-leg hopping,¹⁰ the chosen method should be explained or referenced in each instance. Accurate calculation of K_{vert} from the double integration method also requires motion of the body to mimic a simple spring-mass model, which may not represent the motion of hopping at lower frequencies (ie, < 2 Hz).

The use of a double integration method for calculation of K_{vert} has the benefit of only requiring a force platform. The capture of 3D motion has the additional benefit of providing the investigator with positional data on the COM that is determined from the weighted average of the modeled segments. In contrast to the double integration method, the segmental analysis method is not limited by the asymmetry of the hop cycle or changes in COM position during the hop cycle. Thus, the segmental analysis method is sensitive to variations in the task and subject, by not modeling the subject as a simple spring mass and likely contributing to the lowest variance for the calculation of K_{vert} and $ACOM_L$. Further, due to the built-in calculations within Visual3D, the simplistic nature of the segmental analysis method and the increased availability and use of 3D motion capture equipment within human movement laboratories, there is a

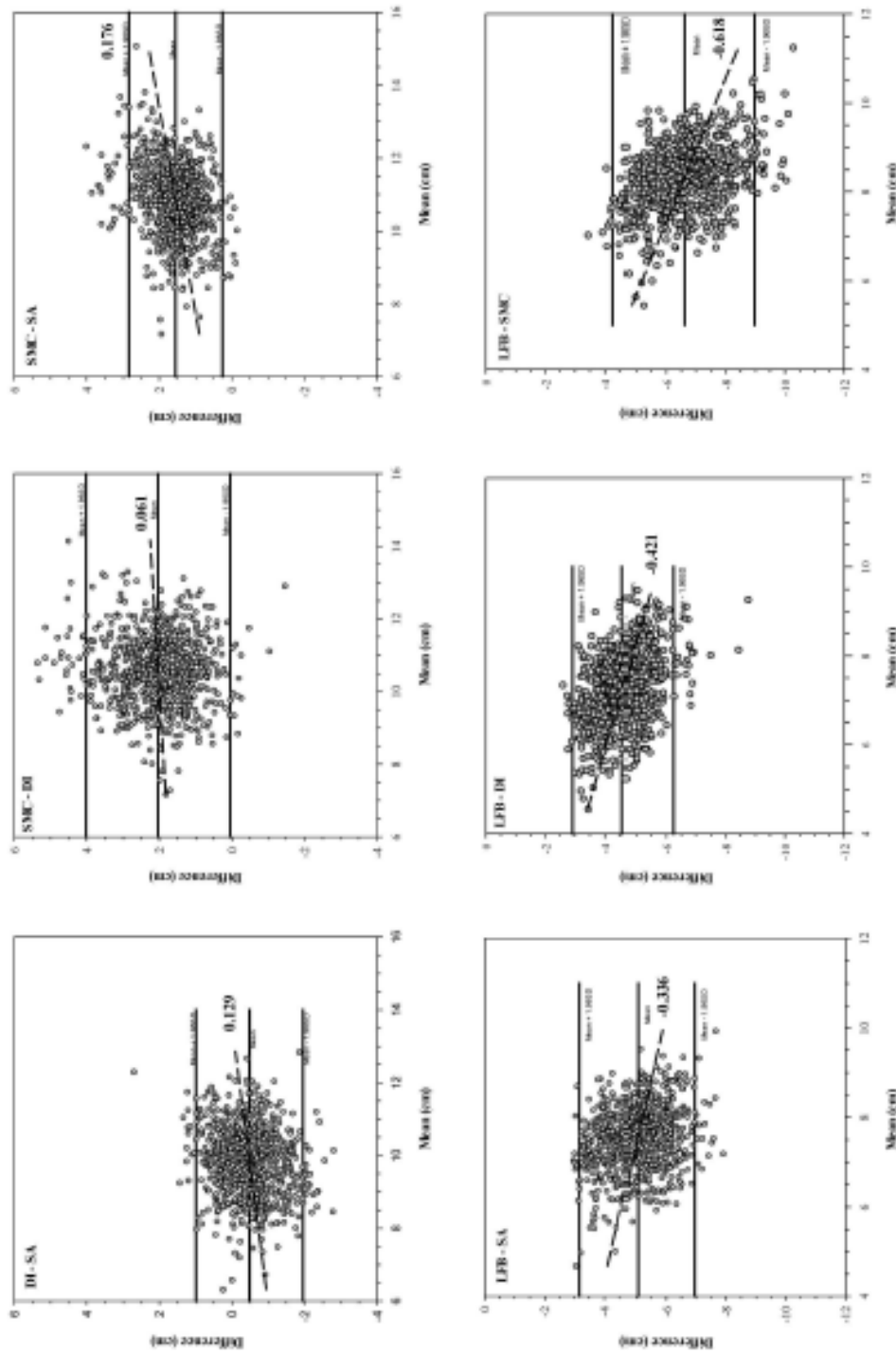


Figure 2 — Bland-Altman plots for the comparison of the vertical displacement of the center of mass (COM) during the loading phase between the sacral marker cluster (SMC), segmental analysis (SA), double integration (DI), and low of falling bodies (LFB) methods. The difference (bias) is represented by the inner horizontal solid line, 95% limits of agreement by the outer solid lines, regression line by the dashed line, and the magnitude of the regression slope labeled to the right. The double integration-segmental analysis method pair demonstrates the smallest bias and trend as vertical displacement of the COM during loading increased.

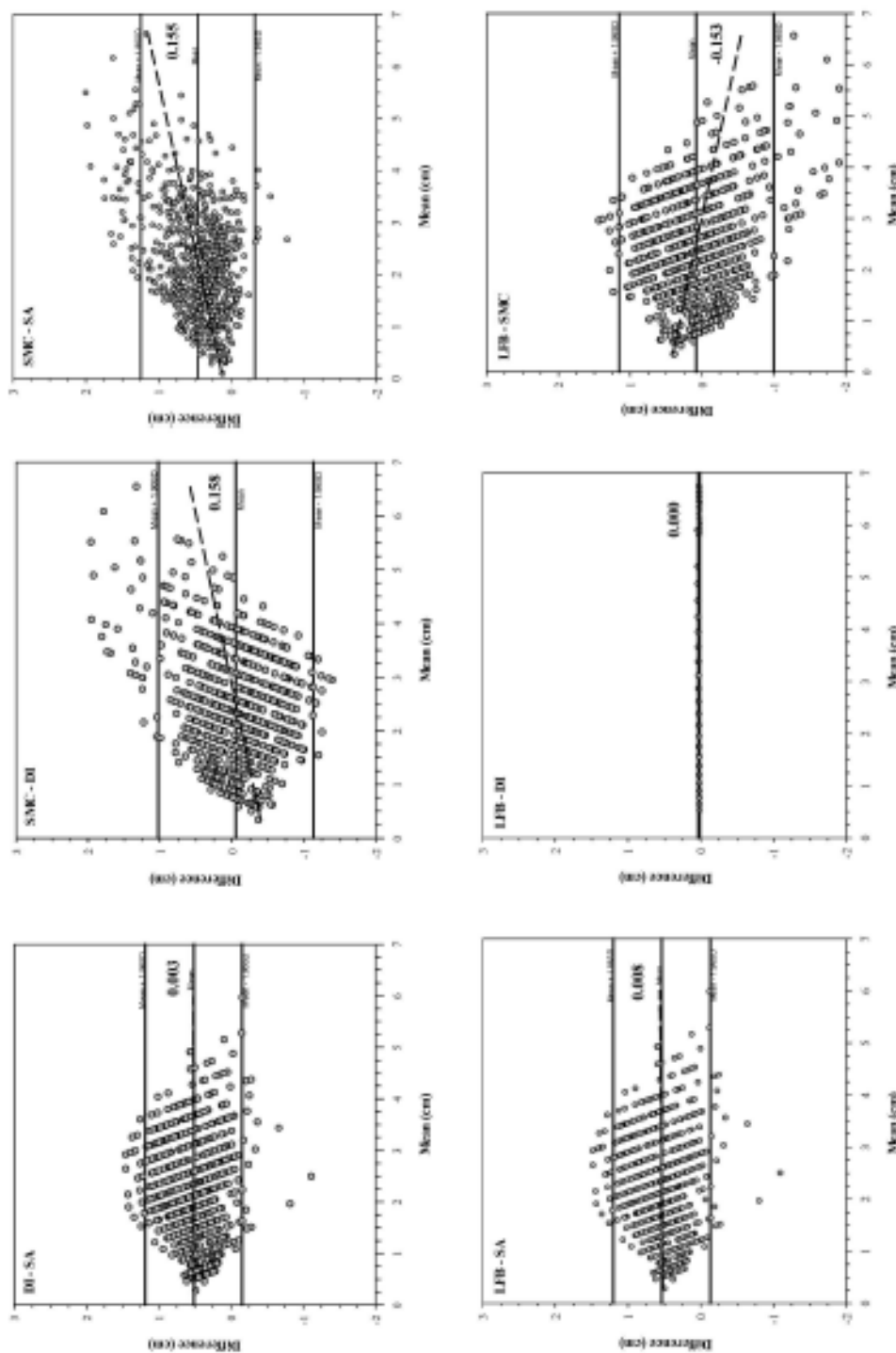


Figure 3 — Bland-Altman plots for the comparison of the vertical displacement of the center of mass (COM) during the flight phase between the sacral marker cluster (SMC), segmental analysis (SA), double integration (DI), and law of falling bodies (LFB) methods. The difference (bias) is represented by the inner horizontal solid line, 95% limits of agreement by the outer solid lines, regression line by the dashed line, and the magnitude of the regression slope labeled to the right. The law of falling bodies—double integration method pair demonstrates the smallest bias. Further, the law of falling bodies—double integration, double integration—segmental analysis, and law of falling bodies—segmental analysis method pairs demonstrate a small trend in vertical displacement of the COM during flight increased.

justifiable reason to use the segmental analysis method as the gold standard for the calculation of K_{vert} during single-leg, on-the-spot hopping. Despite the assumptions of the double integration method, it maintained a good agreement with the more sophisticated segmental analysis method for calculating K_{vert} and should be considered appropriate if there is only force plate data available.

Calculation of K_{vert} was consistently lower using the sacral marker cluster method due to higher values of ΔCOM_L . The positive bias observed on Bland-Altman plots highlight a proportional systematic error between the sacral marker cluster and the other 3 methods when calculating K_{vert} , ΔCOM_P , and ΔCOM_L . In contrast to the segmental analysis method, the sacral marker cluster method assumes a stationary COM position and cannot detect changes in COM position when other body segments move. Further, when using a direct measure as an estimate of the COM position, confounding factors such as pelvic tilt and movement of the sacral rigid body on the skin during the hopping task may overestimate movement of COM position. These errors may contribute to the relative higher values of the vertical displacement of the COM and proportional systematic error.

Although there was a smaller bias between the sacral marker cluster and other methods when the measure of ΔCOM_P and ΔCOM_L were relatively small in magnitude, this is most likely due to the minimization of the errors when there is relatively less displacement and movement of the body's segments. Therefore, the sacral marker cluster may be an appropriate method for the calculation of vertical displacement of the COM for single-leg tasks with relatively less displacement, such as when hopping at higher frequencies.^{12,16} In contrast, there was a greater bias between the sacral marker cluster and all other methods when K_{vert} was greater. Therefore, the sacral marker cluster may not be appropriate for the calculation of K_{vert} during situations where K_{vert} is relatively high, such as on soft surfaces.

The law of falling bodies method demonstrated the least agreement between each pair of methods for the calculation of K_{vert} and ΔCOM_L . Similar to the sacral marker cluster, a proportional systematic error was revealed between the law of falling bodies method when compared with all other methods for the calculation of K_{vert} and ΔCOM_L . Further, the law of falling bodies method had the greatest variance for the calculation of K_{vert} and ΔCOM_L compared with all other methods. Calculation of K_{vert} and ΔCOM_L may not be accurate using the law of falling bodies method due to a number of assumptions which may not be true. The law of falling bodies method assumes maximum vertical velocity of the COM to occur at initial contact (IC) followed by a linear decrease to peak vGRF at which moment there is assumed to be a change in direction and thus velocity to be $0 \text{ m}\cdot\text{s}^{-1}$. However, this has been shown to be inaccurate with a report that maximum vertical velocity of the COM occurs when vGRF is equal to body mass⁸ during the loading phase. Therefore, calculated vertical velocity of the COM during loading will be lower, leading to a lower value for ΔCOM_L and higher K_{vert} value.

In contrast, the law of falling bodies and double integration method demonstrated the greatest agreement for the calculation of ΔCOM_P . The near identical results (bias = -0.02 cm) between methods are due to the same underlying principle of acceleration due to gravity being constant during the flight phase. The difference between the 2 methods being that the double integration method calculates ΔCOM_P on a point-by-point basis while the law of falling bodies method calculated ΔCOM_P as a mean value over the second half of the flight phase.

In conclusion, the segmental analysis method is considered to represent the gold standard measure of K_{vert} during single-leg, on-the-spot hopping. The calculation of K_{vert} from the segmental analysis method minimizes the number of assumptions and has the highest repeatability (lowest variance) across multiple measurements. The double integration method is most comparable to the segmental analysis method and can be used for the accurate estimation of K_{vert} , ΔCOM_L , and ΔCOM_P during single-leg hopping. Care should be taken when comparing K_{vert} , ΔCOM_L , and ΔCOM_P computed from the sacral marker cluster to other methods. The law of falling bodies method should be avoided for the calculation of K_{vert} and ΔCOM_L without first determining a valid conversion factor.

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Appendix B: Mudie et al., (2016). Journal article published in Human Movement Science

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Adaptation of lower limb movement patterns when maintaining performance in the presence of muscle fatigue



Kurt L. Mudie*, Amitabh Gupta, Simon Green, Peter J. Clothier

School of Science and Health, Western Sydney University, Sydney, NSW, Australia

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ABSTRACT

Adaptations in lower limb movement patterns were examined when performance was maintained during a fatiguing repetitive loading task. Forty recreationally active male and female participants performed single-leg hopping to volitional exhaustion at 2.2 Hz to a submaximal height. Spatio-temporal characteristics, mechanical characteristics and variability of the knee-ankle and hip-knee joint couplings were determined at 20% increments during the duration of the hopping task. Variability of the knee-ankle and hip-knee couplings in the flexion/extension axis significantly increased during the loading and propulsion phases during the hopping task ($p < 0.05$). Performance (vertical stiffness, hopping frequency and height) did not change significantly during the task ($p > 0.05$), however foot contact time increased progressively during this task ($p < 0.05$) and maximum hop height significantly decreased after the task ($p < 0.05$). The observed increase in variability between adjoining lower limb segments demonstrated the ability of the neuromotor system to adapt and maintain performance even with the onset of fatigue. This finding highlights that during the performance of a rapid and repetitive loading activity, performance can be preserved when there is variability in the neuromotor system.

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1. Introduction

Variability is an inherent characteristic of human movement that occurs at multiple levels of movement organisation (Preatoni et al., 2013). Although low variability in performance output is desirable, variability in the movement between couplings (two joints or segments) (Fig. 1) has been suggested to play a functional role and contribute to a successful performance output during repetitive tasks (Hamill, Palmer, & van Emmerik, 2012; Latash, 2012; Preatoni et al., 2013). Coupling variability may provide flexibility to the system by permitting adaptation to movement errors or changes in intrinsic or extrinsic factors, such as fatigue or the environment respectively (Bartlett, Wheat, & Robins, 2007; Hamill et al., 2012; Hamill, van Emmerik, Heiderscheit, & Li, 1999; Preatoni et al., 2013). Therefore, greater coupling variability is postulated to be beneficial by permitting multiple movement solutions to a specific task (Bartlett et al., 2007; Hamill et al., 2012, 1999; Preatoni et al., 2013). However, too much or too little coupling variability may be detrimental to the musculoskeletal system and associated with pathology whereby, optimal variability is within the range of these extremes (Bartlett et al., 2007; Hamill et al., 2012, 1999; Preatoni et al., 2013).

There are conflicting findings in the literature of changes in coupling variability as fatigue progresses during repetitive tasks. Coupling variability of the thigh-shank and the shank-foot did not change during running to fatigue (Miller,

* Corresponding author at: Western Sydney University, Campbelltown Campus, School of Science & Health, Building 24, Level 2, Room 100, Locked Bag 1797, Penrith, NSW 2751, Australia.

E-mail address: K.Mudie@westernsydney.edu.au (K.L. Mudie).

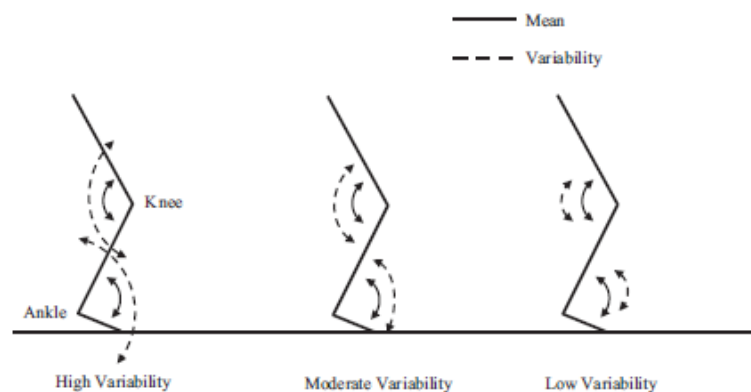


Fig. 1. Schematic diagram representing mean (solid line) movement for each joint of the knee – ankle coupling over a consecutive number of cycles with the variability (dashed line) overlaid. Although mean joint movement remains the same, coupling variability is high on the left, moderate in the middle and low on the right.

Meardon, Derrick, & Gillette, 2008). In contrast, variability of the hip-knee coupling decreased during a 45° anticipated cutting task following an isolated hamstring fatigue protocol (Samaan, Hoch, Ringleb, Bawab, & Weinhandl, 2015). Further, variability of the shank-rear foot coupling was shown to increase during treadmill walking following localised fatigue of the tibialis posterior muscle (Ferber & Pohl, 2011). Trunk-thigh and thigh-shank coupling variability also increased during the performance of a repetitive maximal vertical jump test (Dal Pupo, Dias, Gheller, Detanico, & Santos, 2013).

These conflicting findings may be due to the measure of variability being sensitive to either, differences in tasks between studies or changes in the performance output that would likely occur during a fatiguing task (Miller et al., 2008). Thus, performance output characteristics that modulate leg stiffness and were not reported in previous studies, such as stride length (Miller et al., 2008), force output (Samaan et al., 2015) or jump height (Dal Pupo et al., 2013), may have changed as fatigue increased and may have directly affected the measurement of coupling variability. Further, it is difficult to differentiate possible effects due to warm-up, motor learning during repetitive tasks or fatigue, when coupling variability was not measured regularly (Dal Pupo et al., 2013; Ferber & Pohl, 2011; Miller et al., 2008; Samaan et al., 2015) during a repetitive task.

It remains unclear as to whether the reported changes in coupling variability were due to changes in task performance or in fact due to fatigue. The purpose of this study was to examine the effect of local muscle fatigue on coupling variability when performance output was maintained during a repetitive loading task. It was hypothesised that coupling variability would increase when performance output was maintained in the presence of increasing muscle fatigue.

2. Methods

2.1. Participants

Forty recreationally active males ($n = 20$) and females ($n = 20$) (mean \pm standard deviation (SD) 22.7 ± 3.0 years of age; 1.7 ± 0.1 m in height; 68.8 ± 10.7 kg in mass) volunteered to participate in this study. All participants were healthy and reported participating in exercise for between 1 and 4 h per week. Participants reported no past or current history of lower limb pathology, injury, pain or lower limb fracture within the six months prior to testing. Participants were excluded if they had a history of lower limb surgery(s). Ethical approval (H1074) was granted by the University of Western Sydney Human Research Ethics Committee and all participants provided written informed consent prior to testing.

2.2. Instrumentation

Kinetic and kinematic data were collected synchronously during single-leg, on-the-spot hopping to volitional exhaustion. Kinetic data were sampled from a multicomponent 600×400 mm force plate (Advanced Mechanical Technology, Inc., model BP400600-1000, Watertown, MA) at 1500 Hz. Kinematic data were sampled using an Optotrak Certus System (Northern Digital Inc., Waterloo, Canada) at 150 Hz (First Principles, Version 1.2.4) and processed using Visual 3D (C-Motion, Version 4, Germantown, MD).

2.3. Participant preparation

Following measurement of each participant's height and body mass, a warm-up and a hopping familiarisation period was completed (Hobara, Kobayashi, Kato, & Ogata, 2013). To indirectly control task performance of vertical stiffness, a target

hopping frequency and height were imposed. Each participant was instructed to land on the audible tone of a metronome set to the preferred hopping frequency for humans at 2.2 Hz (132 beats/minute) that was consistent with previous studies (Farley, Blickhan, Saito, & Taylor, 1991; Hobarra, Kobayashi, et al., 2013). Tactile feedback was provided to allow each participant to hop to a predetermined target height by a custom-built apparatus on which 32 mm wide elastic bands were stretched horizontally above the participant's head. All trials were performed barefoot (Hobarra, Inoue, Omuro, Muraoka, & Kanosue, 2011; Zuur et al., 2010) and on the participant's dominant leg, defined as their preferred kicking leg (Hobarra, Inoue, & Kanosue, 2013; Padua et al., 2006). Each participant was instructed to keep their hands on their hips, land on the audible tone of the metronome on the ball of their foot and lightly touch their head on the elastic bands when hopping to the target height.

A seven-segment model was used to model the trunk (Wu, van der Helm, et al., 2005), pelvis (Bell, Pedersen, & Brand, 1990), thigh, shank (Ball, 2011), hindfoot, forefoot (Wu, Su, et al., 2000) and hallux (Stebbins, Harrington, Thompson, Zavatsky, & Theologis, 2006) of the hopping leg. Twenty-one active markers were adhered over the segments, including a single four-marker cluster (sacral), three three-marker clusters (T8, distal lateral thigh and shank) and eight individual active markers (foot) (Cappozzo, Cappello, Croce, & Pensalfini, 1997; Stagni, Fantozzi, Cappello, & Leardini, 2005). Fifteen calibration markers were digitised to identify the C7 and T8 spinous process, suprasternal notch, xiphoid process, left and right acromioclavicular joints, left and right anterior superior iliac spines, left and right posterior superior iliac spines, greater trochanter, medial and lateral femoral epicondyles and medial and lateral malleoli of the dominant leg. Eight individual active markers (NDI, Optotrak, Canada) (7 mm diameter) were adhered to the posterior, medial and lateral aspects of the calcaneus, base and head of the first and fifth metatarsals and the medial aspect of the base of the proximal phalanx of the great toe.

2.4. Controlling task performance

Target hop height was calculated for each participant by recording a trial of single-leg hopping at a cadence of 2.2 Hz for 15 s. Vertical displacement of the centre of mass (COM) during the flight phase was calculated for each hop cycle (Ranavolo et al., 2008) (see Sections 2.6–2.7.1 for detailed description of the processing and analysis). Target hop height was set to each participant's preferred hopping height at 2.2 Hz, calculated as the mean vertical displacement of the COM during the flight phase from the sixth to the tenth hop cycle (Hobarra, Inoue, et al., 2013; Hobarra, Kobayashi, et al., 2013). The elastic bands were then set at the target hop using a laser measurement device (Bosch PLR 50, measurement accuracy ± 2 mm).

2.5. Testing protocol

A static calibration trial with the participant standing in an anatomical neutral position was recorded. Each participant then completed a single effort of on-the-spot single-leg hopping to volitional exhaustion at 2.2 Hz to a target hop height. Volitional exhaustion was defined as the moment when the participant could no longer maintain the required performance output of hopping frequency or target hop height (Morio et al., 2011; Nicol, Avela, & Komi, 2006; Regueme, Nicol, Barthélemy, & Grélot, 2005). To quantify the acute functional effects at volitional exhaustion, five continuous maximal effort single-leg hops were performed three minutes before (PRE) and 10 s after (POST) the exhaustive single-leg hopping trial.

2.6. Data processing

All recorded trials were exported from First Principles software as C3D files for processing. Kinematic marker recordings were visually inspected to determine missing frames and missing data were interpolated using spline interpolation for up to a maximum gap of 10 frames. A third order polynomial equation was used to fill the gap and the three frames before and after the missing frame(s) used to calculate the coefficient of the polynomial equation. Recorded marker positions and force plate data were filtered using a fourth order zero-lag bidirectional Butterworth filter with an 8 Hz cut-off (Bobbert & Richard Casius, 2011; Hobarra et al., 2011; Hobarra, Kimura, Omuro, Gomi, Muraoka, Iso, et al., 2008) and a fourth order bidirectional low-pass Butterworth filter with a 50 Hz cut-off (Gupta, Hilliard, Mudie, & Clothier, 2016; Gupta, Mudie, & Clothier, 2014), respectively.

Joint axes were defined by the right-handed Cartesian local coordinate system for two adjacent body segments to determine the positive or negative direction of motion (Wu et al., 2002). Motion of the segments were expressed as the orientation of the distal segment with reference to the neighbouring proximal segment (Doets, van Middelkoop, Houdijk, Nelissen, & Veeger, 2007; Farris & Sawicki, 2012; Stebbins et al., 2006; Wu et al., 2002), and the medio-lateral (y) and superior–inferior (z) axes mirrored in the left limbs to ensure consistency between sides regarding positive/negative motion about a joint axis (Grood & Suntay, 1983).

2.7. Data analysis

A single hop cycle was defined as a complete flight phase and subsequent contact phase. The contact phase included the loading (initial contact to peak vertical ground reaction force (vGRF)) and propulsive (peak vGRF to toe-off) phase, determined from the vGRF trace (Lloyd, Oliver, Hughes, & Williams, 2009). Contact time represented the time the foot was in contact with the force plate and was calculated as the duration between initial contact and toe-off for each hop cycle

(Lloyd et al., 2009). Dependant variables calculated for each hop cycle during the exhaustive trial included hopping frequency, hopping height, contact time, vertical stiffness and coupling variability of the knee-ankle and hip-knee couplings. Dependant variables were calculated as the mean of 10 consecutive hop cycles (Kuitunen, Ogiso, & Komi, 2011; Maton & Pellec, 2001; Padua et al., 2006) at time periods of 0, 20, 40, 60, 80 and 100% of the duration of the trial (Microsoft Office Excel, 2007 and MATLAB, 2012B 32-bit). To quantify the acute functional effects at volitional exhaustion, the single-hop cycle with the greatest hop height was chosen during the PRE and POST maximal effort hop trials.

2.7.1. Performance output

Temporal events including toe-off, initial contact and peak vertical ground reaction force (vGRF) were identified for each hop cycle (Lloyd et al., 2009). Maximal vertical displacement during flight and contact were determined by a segmental analysis method (Ranavolo et al., 2008). Vertical stiffness (k) was calculated as the quotient of force normalised to body mass and displacement during the loading phase (Eq. (1)) (Butler, Crowell, & Davis, 2003; Farley, Houdijk, van Strien, & Louie, 1998; Farley & Morgenroth, 1999; Ferris & Farley, 1997; Serpell, Ball, Scarvell, & Smith, 2012).

$$k = \text{peak vGRF}/z_i \tag{1}$$

where k was vertical stiffness ($\text{N kg}^{-1} \text{m}^{-1}$), peak vGRF was the maximum vGRF during the contact phase normalised to body mass (N kg^{-1}) and z_i was the loading height (m).

2.7.2. Joint coupling variability

Intra-limb joint coupling variability is a linear tool that investigates the magnitude of movement variability (Hamill et al., 2012; Preatoni et al., 2013), calculated to describe the variability in coordination between the knee and ankle joints and hip and knee joints. Joint coupling variability was calculated for knee flexion/extension – ankle flexion/extension (KxAx), hip flexion/extension – knee flexion/extension (HxKx) and knee flexion/extension – ankle eversion/inversion (KxAy) during contact with the force plate. These couplings were chosen due to the predominant unidirectional motion of hopping, the role of the ankle and knee joints for modulating submaximal hopping (Lamontagne & Kennedy, 2013) and the kinetic chain between lower limb joints. Quantification of coupling variability was calculated using a modified vector coding technique (Ferber, Davis, & Williams, 2005; Heiderscheit, Hamill, & van Emmerik, 2002; Pollard, Heiderscheit, Van Emmerik, & Hamill, 2005). Initially all joint angle data during the hop cycle loading and propulsive phases were normalised to 101 data points (Ferber et al., 2005; Heiderscheit et al., 2002; Pollard et al., 2005). Motion of the proximal and distal joints were then plotted on the ordinate and abscissa, respectively (Ferber et al., 2005). The coupling angle (θ) was calculated as the orientation of the resultant vector to the right horizontal between two adjacent data points as follows (Fig. 2; Eq. (2)):

$$\theta_i = \text{abs}[\tan^{-1}(y_{i+1} - y_i/x_{i+1} - x_i)] \tag{2}$$

where i = data point 1, 2 and n of the time series (Ferber et al., 2005; Heiderscheit et al., 2002; Pollard et al., 2005). Calculated values were then converted from radians to degrees with the resultant coupling angles ranging from 0° to 90° (Ferber et al., 2005; Pollard et al., 2005). A coupling angle of 45° indicated equal motion from both proximal and distal joints. A coupling angle greater or less than 45° indicated greater motion in the proximal or distal joint than the other joint, respectively (Ferber et al., 2005). To calculate the mean joint coupling angle across the loading and propulsion phases, the mean coupling angle between each data point was calculated on a point-by-point basis across 10 consecutive hop cycles, resulting in a mean for each of the 100 time points over each phase (Fig. 3) (Ferber et al., 2005; Heiderscheit et al., 2002; Pollard et al., 2005). At each data point, the SD of the coupling angle across the 10 hop cycles was calculated, resulting in a 100 point variability time

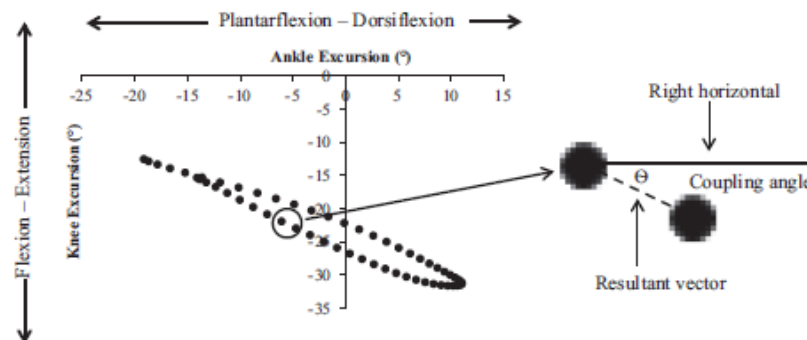


Fig. 2. Vector coding technique used to calculate coupling angles (θ) between two joints. Plotted knee and ankle excursions during the contact phase (left) and an enlarged image of two subsequent data points with θ indicated (right).

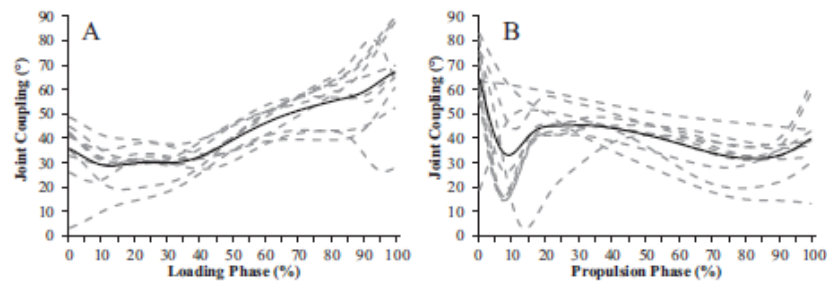


Fig. 3. Knee-ankle joint coupling ($^{\circ}$) as a function of the (A) loading and (B) propulsion phase (0–100%) for a single representative male participant. The 10 dashed light lines represent the calculated joint coupling angle for the first 10 hop cycles of the trial. The single solid dark line represents the calculated mean joint coupling angle of the first 10 hop cycles as a function of the (A) loading and (B) propulsion phase (0–100%).

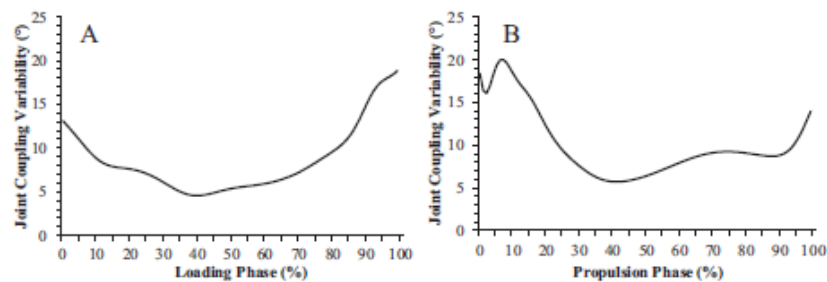


Fig. 4. Knee-ankle joint coupling variability ($^{\circ}$) as a function of the (A) loading and (B) propulsion phase (0–100%) for a single representative male participant. Each time point (0–100%) was calculated as the standard deviation of the coupling angle at that time point across the first 10 hop cycles, resulting in a 100 point variability time series across the (A) loading and (B) propulsion phase, providing a measure of between-cycle, within-participant variability.

series for the loading and propulsion phases (Fig. 4) (Ferber et al., 2005; Heiderscheidt et al., 2002; Pollard et al., 2005). The mean joint coupling variability during the loading and propulsion phases was calculated as the mean value of the variability time series during each phase, providing a measure of between-cycle, within-participant variability (Fig. 4) (Ferber et al., 2005; Heiderscheidt et al., 2002; Pollard et al., 2005). Joint coupling variability from 10 consecutive hop cycles was determined for the loading and propulsive phases at 0, 20, 40, 60, 80 and 100% of the trial (Ferber et al., 2005; Heiderscheidt et al., 2002; Pollard et al., 2005).

2.8. Statistical analyses

A one-way (time period) ANOVA with repeated measures (0, 20, 40, 60, 80 and 100% time periods) was performed to compare all dependant variables during the exhaustive trial (SPSS, Version 22). Mauchly's test of sphericity was performed and if violated ($p < 0.05$) a Greenhouse-Geisser procedure used. *Post-hoc* pairwise multiple comparisons were made between the 0% time period and subsequent time periods with a Bonferroni correction applied. To compare maximal hop height between PRE and POST trials a two-tailed paired sample *t*-test was performed. Alpha level set *a priori* at $p < 0.05$ for all statistical analyses.

3. Results

3.1. Fatigue and performance output

Total duration of single-leg hopping (mean \pm SD) to volitional exhaustion was 79 ± 27 s. The target hop height for all participants was 2.50 ± 1.08 cm (range 0.65–5.00 cm). There was a significant decrement ($p < 0.001$) of $22 \pm 16\%$ in maximum hopping height (PRE: 9.38 ± 2.81 cm; POST: 7.23 ± 2.49 cm) following the exhaustive hopping trial. There was a significant increase in contact time during the exhaustive hopping trial from the 0% time period to the 80% ($p = 0.044$) and 100% ($p = 0.007$) time periods (Fig. 5). In contrast, there was no significant change detected for performance output characteristics, including hopping frequency, hopping height and vertical stiffness between each time period (Table 1).

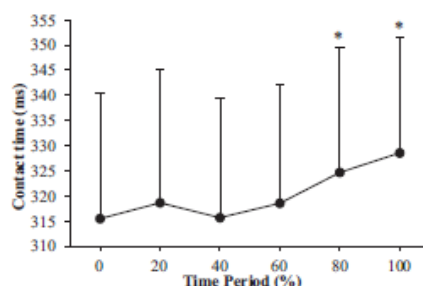


Fig. 5. Mean (\pm SD) contact time (ms) at each time period over the duration of hopping. *Indicates $p < 0.05$ compared to the 0% time period.

Table 1
Performance output characteristics (mean (SD)) at all time periods during the exhaustive single-leg hopping trial.

	0%	20%	40%	60%	80%	100%
Hopping frequency (Hz)	2.23 (0.05)	2.21 (0.03)	2.22 (0.04)	2.23 (0.04)	2.21 (0.05)	2.23 (0.05)
Hopping height (cm)	1.82 (0.85)	1.83 (0.96)	1.90 (0.84)	1.82 (0.84)	1.80 (0.88)	1.71 (0.77)
Vertical stiffness ($\text{N kg}^{-1} \text{m}^{-1}$)	247.88 (23.48)	243.11 (21.88)	245.41 (20.20)	244.20 (23.16)	238.51 (25.14)	237.22 (21.70)

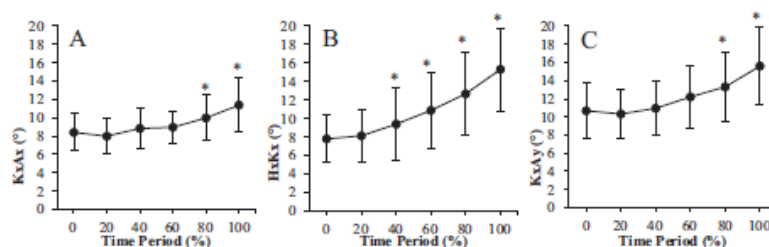


Fig. 6. Mean (\pm SD) coupling variability ($^{\circ}$) during loading at each time period for: (A) knee flexion/extension – ankle flexion/extension (KxAx); (B) hip flexion/extension – knee flexion/extension (HxKx); and (C) knee flexion/extension – ankle abduction/adduction (KxAy). *Indicates $p < 0.05$ compared to the 0% time period.

3.2. Joint coupling variability

During the loading phase, there was a significant increase in KxAx and KxAy between the 0% time period and the 80% ($p = 0.002$; $p = 0.005$) and 100% ($p < 0.001$; $p < 0.001$) time periods. There was a significant increase in HxKx during the loading phase between the 0% time period and the 40% ($p = 0.011$), 60% ($p < 0.001$), 80% ($p < 0.001$) and 100% ($p < 0.001$) time periods (Fig. 6).

During the propulsion phase, there was a significant increase in KxAx between the 0% time period and the 100% ($p < 0.001$) time period, and HxKx between the 0% time period and the 80% ($p = 0.002$) and 100% ($p < 0.001$) time periods. In contrast, no significant difference was detected for KxAy during propulsion between the 0% time period and any subsequent time period (Fig. 7).

4. Discussion

This study examined adaptations in joint coupling variability when performance output was maintained as local muscle fatigue increased during a repetitive loading task. The results of the current study supported the alternate hypothesis, that there was an increase in joint coupling variability when performance output was maintained with increasing muscle fatigue. This suggests that during a fatiguing rapid and repetitive loading task, healthy individuals were able to maintain a required performance level with multiple movement patterns.

The single-leg hopping protocol to exhaustion was considered to be an adequate stimulus to elicit an increase in muscle fatigue of the lower limb muscles. This was supported by the finding of a significant reduction in post-trial maximum

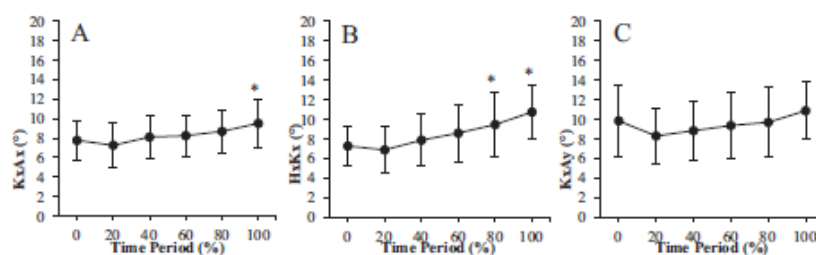


Fig. 7. Mean (\pm SD) coupling variability ($^{\circ}$) during propulsion at each time period for: (A) knee flexion/extension – ankle flexion/extension (KxAx); (B) hip flexion/extension – knee flexion/extension (HxKx); and (C) knee flexion/extension – ankle abduction/adduction (KxAy). *Indicates $p < 0.05$ compared to the 0% time period.

hopping height and an increase in contact time during the trial performance, consistent with previous research on the effect of fatigue on a repeated jump task (Dal Pupo et al., 2013; McNeal, Sands, & Stone, 2010; Morio et al., 2011; Morio, Nicol, Barla, Barthélemy, & Berton, 2012).

A key difference between previous studies and the current study was that the task performance was maintained such that vertical stiffness remained unchanged throughout the entire trial by imposing a target hopping cadence and height (Farley et al., 1991; Ferris & Farley, 1997; Hobarata et al., 2011). This was achieved despite the onset of and increasing fatigue during the trial. The capacity to withstand increasing fatigue and maintain performance could be explained by the notion that limb multi-segment coordination may change even though the task remains unchanged. Inherent to this is evidence that local muscle fatigue of the triceps surae does lead to a reduced ability to generate maximal power output (Dal Pupo et al., 2013). Due to the multi-segment coordinative strategies available, it is plausible that there is an inability to generate adequate ankle plantarflexion moment during contact, such that there were changes in the contribution of the thigh and hip muscles which produced compensatory movements. It is speculated that these compensatory movements accommodated the required ankle joint power when the ankle plantarflexors were working both eccentrically during the loading phase and concentrically during the propulsive phase to maintain performance.

Lower limb compensatory movements are reflective of changes in variability in the movement between two joints or segment couplings. Findings in the current study demonstrate significant increases in HxKx and KxAx coupling variability towards the end of the trial when fatigue was more evident, indicated by a greater contact time. The current results support previous findings of increased hip-knee and knee-ankle coupling variability when fatigued (Dal Pupo et al., 2013; Ferber & Pohl, 2011).

In the current study there were increases in coupling variability during both the loading and propulsive phases. In contrast to previous results during a repetitive maximal vertical jump test (Dal Pupo et al., 2013), the current study demonstrated increases in coupling variability during the loading phase and were of higher magnitude than during the propulsive phase (Figs. 6 – 7). The contrasting results may be due to the task being submaximal in the current study compared to the maximal effort task by Dal Pupo et al. (2013). As fatigue progresses during a maximal effort task, changes are suggested to occur more in characteristics that directly impact performance, such as during propulsion to achieve a maximal jump height (Nicol et al., 2006). However, during a submaximal task, compensatory strategies that allow performance to be maintained are suggested to be more evident, such as during loading (Nicol et al., 2006). Thus, it is plausible that the observed increase in variability during the current submaximal task was a compensatory strategy that may be used to withstand the effects of fatigue and allow maintenance of the required performance output. Further, maintaining performance despite an increase in coupling variability during the loading phase may also support the functional role of coupling variability during repetitive loading tasks.

The observed increase in coupling variability during loading may be explained by the greater tension of lower limb musculature that assists to decelerate the body in preparation for the upward propulsive phase (Cavagna, 1977). In contrast to the loading phase, there may be lower levels of muscle activity during the propulsive phase with a greater reliance on utilising stored elastic energy (Lamontagne & Kennedy, 2013). Thus, greater stress on lower limb musculature can be experienced attenuating the landing forces compared to the upward propulsion. Increased coupling variability during the loading phase may allow the landing forces to be more optimally attenuated by distributing the load across various structures and/or muscles (Hamill et al., 1999; Miller et al., 2008). Further, increased coupling variability may permit flexibility to the lower limb to adapt to increases in local muscle fatigue, which may affect the landing (Pollard et al., 2005; Wilson, Simpson, van Emmerik, & Hamill, 2008).

In addition to the lower increase in coupling variability during the propulsive phase, the increase in the variability of the knee-ankle couplings was also lower in magnitude than the increase in hip-knee coupling variability. Specific to the current task, joint motions of predominantly the ankle and knee during the propulsive phase were characteristics that were observed to modulate submaximal hopping height (Dal Pupo et al., 2013; Lamontagne & Kennedy, 2013). With increasing fatigue of the lower limb muscles, the increase of knee-ankle coupling variability was not associated with a change in performance.

Although coupling variability is beneficial during repetitive tasks, too much variability in characteristics that modulate performance may cause the motor system to be unstable and affect the ability to achieve the required performance (Hamill et al., 2012). Therefore, increases in coupling variability may be limited by the constraints of the task and whether the task is able to be maintained at a constant output. Thus, as the task of hopping was predominantly modulated by the knee-ankle coupling, a relatively large magnitude of variability in the hip-knee coupling may allow enough source of variability in the kinetic chain and still maintain a consistent performance.

As the task progressed variability of the HxKx coupling increased, with a significant increase occurring early in the trial from the 40% time period onwards. A subsequent and significant increase in the KxAx coupling occurred onwards from the 80% period. In a task where there is a required performance output goal, a change in control of one element will likely require a reactive compensation from another in an attempt to maintain the required goal of the movement (Latash, Levin, Scholz, & Schöner, 2010). Thus, the latter significant increase in variability of the KxAx coupling may be a reactive compensatory mechanism to both increased muscle fatigue and to maintain the performance output of the task. The current findings support previous research, highlighting that compensatory mechanisms are observed during other fatiguing and repetitive tasks when maintaining performance such as finger tapping (Singh, Varadhan, Zatsiorsky, & Latash, 2010), postural sway (Singh & Latash, 2011) and walking (Ferber & Pohl, 2011). The results of the current study reveal that appropriate compensatory strategies are also present during a rapid and repetitive loading task in the lower limb.

5. Conclusion

Increased coupling variability may allow greater available movement solutions between the coupled joints as muscle fatigue progresses. This may be an effective strategy to overcome the effects of increased local muscle fatigue by permitting alternate movement patterns with the maintenance of motor performance. Alternate movement patterns which would distribute the load across various muscles may delay the detrimental effects of fatigue and preserve performance output. This finding supports the notion that performance is able to be preserved despite the presence of variability in the neuromotor system.

Conflict of interest disclosure

The authors declare no conflict of interest.

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**Appendix C: Mudie et al., (2015). Conference abstract presented
at the 33rd International Conference on Biomechanics in
Sport**

**DOES THE METHOD OF MEASURING CENTRE OF MASS
DISPLACEMENT AFFECT VERTICAL STIFFNESS CALCULATION IN
SINGLE-LEG HOPPING?**

Kurt L. Mudie, Amitabh Gupta and Peter J. Clothier

**School of Science and Health, University of Western Sydney,
Sydney, NSW, Australia**

The purpose of this study was to compare vertical stiffness values calculated from two kinetic and two kinematic estimations of the vertical displacement of the centre of mass. Twenty recreationally active male and female participants completed one 15 s single-leg hopping trial at 2.2 Hz with vertical stiffness calculated for the first 10 complete hop cycles. Vertical displacement was estimated using double integration (DI), first principle (FP), sacral marker cluster (SMC) and segmental analysis (SA) methods. Bland-Altman plots demonstrated the SA and DI methods to have a small bias (0.92 kN/m) and tight 95% limits of agreement (-1.16 to 3.08 kN/m). In contrast, the SMC and FP methods underestimated and overestimated vertical stiffness, respectively. These findings suggest the SA and DI methods can be used interchangeably to calculate vertical stiffness.

KEY WORDS: agreement, lower limb, Bland-Altman, double integration.

INTRODUCTION: Lower limb stiffness is often measured during tasks of locomotion such as walking, running and hopping and has been suggested to influence a number of performance and injury risk characteristics (Butler et al., 2003). Thus a consistent and accurate calculation of lower limb stiffness is of interest to many researchers and sports practitioners. Specifically, during on-the-spot-hopping vertical stiffness is calculated as the quotient of maximal ground reaction force and vertical centre of mass (COM) displacement (Butler et al., 2003). Ground reaction forces can be directly measured from a force platform, however it is not possible to measure the exact position of the COM. A number of methods have been developed to estimate vertical displacement of the COM, each with specific advantages and disadvantages (Hébert-Losier & Eriksson, 2014; Hobara et al., 2014; Ranavolo et al., 2008). However the accuracy of different methods for the calculation of vertical displacement may be task dependent (Ranavolo et al., 2008). Therefore, it remains unknown whether vertical COM displacements derived from different methods provides similar calculations of vertical stiffness during single-leg hopping. This remains problematic for the interpretation and comparison of new and existing research. The purpose of this study was to compare vertical stiffness values calculated from using two kinetic and two kinematic estimations of the vertical displacement of the COM during single-leg hopping.

METHODS: Following a warm-up and familiarisation period (Hobara et al., 2011), twenty healthy recreationally active male and female participants completed 15 s of on-the-spot single-leg hopping on a force platform. Hopping was performed barefoot on the participants self-selected dominant leg (Padua et al., 2006) at 2.2 Hz controlled by an audible digital metronome. Kinetic (AMTI, Gen 5, USA) and kinematic data (NDI,

Optotrak, Canada) were collected synchronously at 1500 Hz and 150 Hz (First Principles software, Version 1.2.4), respectively. Consistent with previous research, a seven-segment model was used to model the trunk (G. Wu et al., 2005), pelvis, thigh, shank (Ball, 2011), hindfoot, forefoot (W. L. Wu et al., 2000) and hallux (Stebbins et al., 2006) of the hopping leg.

Kinetic data were dual-pass filtered with a low pass Butterworth filter with a 50 Hz cut-off. Kinematic data were interpolated using a spline interpolation for up to a maximum gap of 10 frames and dual-pass filtered using a fourth order Butterworth filter with an 8 Hz cut-off (Hobara et al., 2011)(Visual 3D, Version 4). Derived variables calculated were vertical stiffness and vertical displacement of the COM during the flight and loading phases. All COM displacement measures were derived using a sacral marker cluster (SMC), segmental analysis (SA), double integration (DI) and first principles (FP) methods as the mean of the first 10 hop cycles that were within 5% of the set hopping frequency (Microsoft Office Excel, 2007).

Vertical stiffness was calculated for each method as the quotient of maximal ground reaction force (N) and vertical displacement of the COM during loading (m) (Butler et al., 2003). The SMC method estimated the vertical displacement of the COM during the flight and loading phases by calculating the vertical displacement of the centre of the sacral marker cluster (Ranavolo et al., 2008). The SA method involved the calculation of the COM of the seven modelled upper and lower limb segments by default within Visual 3D (Version 4) using the location and masses of each segment. Vertical displacement of the COM was then calculated during the flight and loading phases. Double integration of the vGRF curve was used to estimate the vertical displacement of the COM during both flight and loading phases for the DI method (Butler et al., 2003; Hébert-Losier & Eriksson, 2014). The FP method used Newton's Laws of motion (Hall, 2007) to estimate the vertical displacement of the COM during the flight and loading phases (equation (1)–(3)). First the vertical displacement of the COM during the falling period of the flight phase (z_f) was calculated as:

$$z_f = (1/2) \times g \times (t_f/2)^2 \quad (1)$$

Following, the velocity of the COM at IC (v_i) was determined by:

$$v_i = (2 \times g \times z_f)^{1/2} \quad (2)$$

Lastly, the vertical displacement of the COM during the loading phase (z_l) was calculated by:

$$z_l = [(v_f + v_i)/2] \times t_l \quad (3)$$

Where g was the acceleration due to gravity (-9.81 m.s^{-2}), t_f was the total time of the flight phase (s), and v_f was the velocity of the COM at peak force (0 m.s^{-1}).

To quantify the agreement between methods, Bland-Altman plots were created by plotting the mean difference (bias) against the mean result of each method pair (Bland & Altman, 1986). The 95% limits of agreement (LoA) were estimated as the mean difference ± 1.96 of the standard deviation of the difference (Bland & Altman, 1999).

RESULTS AND DISCUSSION: The main finding of this investigation revealed the SA and DI methods produced similar calculations of vertical stiffness during single-leg hopping. Bland-Altman plots revealed only a small bias (0.92 kN/m) and tight 95% LoA (-1.16 to 3.08 kN/m) between the SA and DI methods for the calculation of vertical stiffness (Figure 1). Further, Bland-Altman plots also revealed no increasing or decreasing trend between the size of the difference (bias) and the mean score of the SA and DI methods. Therefore, the current results suggest the SA and DI methods can be used interchangeably for the calculation of vertical stiffness during single-leg hopping. Although the DI method only requires a force platform and thus use outside

of the laboratory is possible, the SA method is able to provide additional data such as three dimensional positions of the COM. Further, due to built-in calculations within common software such as Visual3D (Visual 3D, Version 4) the simplistic nature of the SA method and the increased availability and use of three dimensional motion capture equipment within human movement laboratories supports the use of the SA method. In contrast, Bland-Altman plots revealed a large bias in vertical stiffness between the SMC and the SA (2.52 kN/m) and DI (3.48 kN/m) methods with an increasing bias trend as mean values increased (Figure 1). Therefore, the SMC method underestimated vertical stiffness with the size of the bias increasing as stiffness values increased and thus is not appropriate for the calculation of vertical stiffness during single-leg hopping at 2.2 Hz. The underestimation of vertical stiffness is due to an overestimation of vertical displacement of the COM which may be caused by a number of factors including pelvic tilt, clothing and skin movement artefact, all of which would be expected to be amplified during tasks with greater movement such as hopping or jumping at lower frequencies. However, the SMC may be an appropriate method for the calculation of vertical stiffness for tasks with less displacement, for example hopping at higher frequencies. Although future research is required to determine whether an increasing trend remains present during tasks when the vertical displacement of the COM is less.

The FP method overestimated vertical stiffness compared to all other methods with Bland-Altman plots revealing a large bias between the FP method and the SMC (21.46 kN/m), SA (18.94 kN/m) and DI (17.98 kN/m) methods (Figure 1). When compared to the SMC, SA and DI methods the FP method also demonstrated an increasing bias as mean values increased. The overestimation of the FP method for the calculation of vertical stiffness may be due to the assumptions required to calculate the vertical displacement of the COM during loading. When using the FP method it is assumed maximum velocity of the COM occurs at IC followed by a linear decrease to zero at peak vGRF. This assumption is incorrect as downward maximum vertical velocity of the COM is not reached until after IC when force equals body weight (Blickhan, 1989) and therefore vertical velocity of the COM during loading will be largely underestimated. Thus compared to other methods that do not rely on this assumption, the FP method will underestimate vertical displacement of the COM during loading causing vertical stiffness to be overestimated.

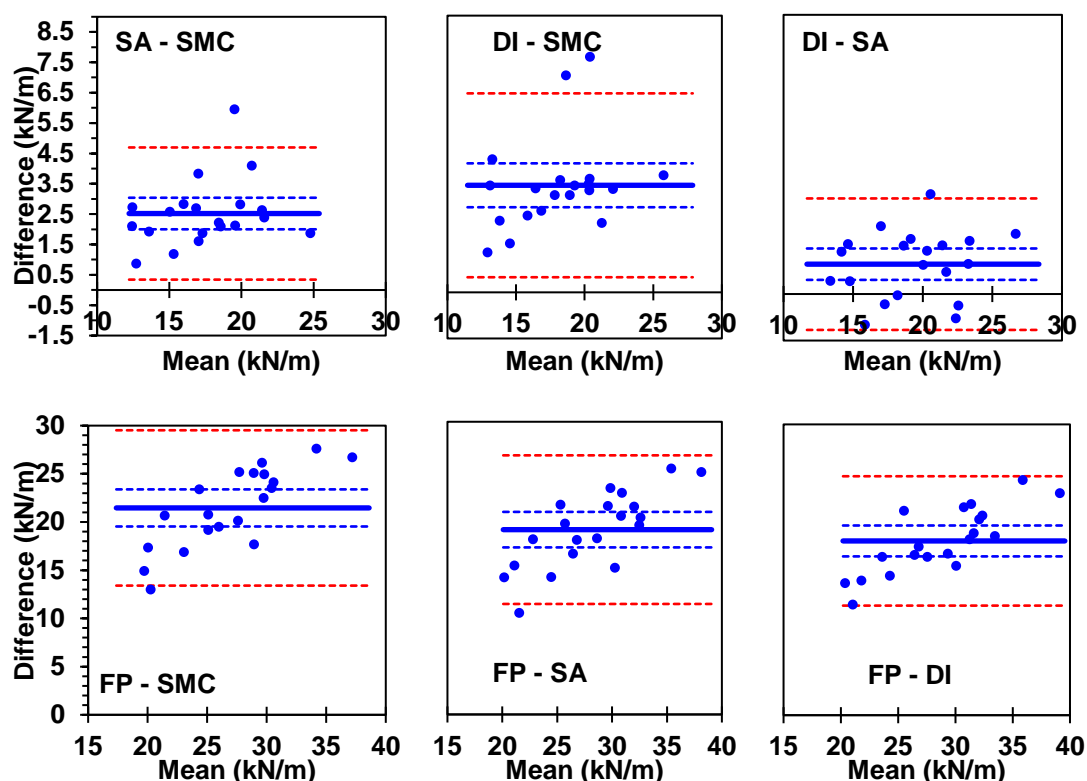


Figure 1: Bland-Altman plots for the comparison of vertical stiffness between the sacral marker cluster (SMC), segmental analysis (SA), double integration (DI) and first principles (FP) methods. Horizontal solid line = mean difference; inside dashed lines = 95% confidence intervals; and outside dashed lines = 95% limits of agreement.

CONCLUSION: This study highlights that the calculation of vertical stiffness is sensitive to the method used to estimate vertical displacement of the COM. This finding has implications for researchers and for interpretation of the scientific literature by practitioners. The current study suggests vertical stiffness calculated from the SA and DI estimations of COM vertical displacement can be appropriately compared and either method used to calculate vertical stiffness during single-leg hopping. However, the SMC and FP methods underestimated and overestimated calculated vertical stiffness values respectively, with the difference between methods increasing as the magnitude of vertical stiffness increased. Due to the invariable bias between the SA and DI methods over a range of vertical stiffness values, the SA and DI methods are suggested to be equally as representative of a measure of vertical stiffness over a greater range of values than the SMC and FP methods when assuming motion to be modelled as a spring-mass. Therefore, it is recommended that either the SA or DI methods be used to calculate vertical stiffness during single-leg hopping at 2.2 Hz.

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**Appendix D: Mudie et al., (2015). Conference abstract presented
at the 33rd International Conference on Biomechanics in
Sport**

**THE EFFECT OF FATIGUE ON LOWER LIMB MOTOR VARIABILITY
DURING A CONTROLLED REPETITIVE STRETCH-SHORTEN CYCLE
TASK**

Kurt L. Mudie, Amitabh Gupta and Peter J. Clothier

**School of Science and Health, University of Western Sydney,
Sydney, NSW, Australia**

This study evaluated changes in lower limb joint coupling variability during single-leg hopping to exhaustion. Twenty recreationally active male and female participants performed single-leg hopping at 2.2 Hz to a target height. At 0, 20, 40, 60, 80 and 100% of the total duration of hopping, spatio-temporal characteristics and variability of the knee-ankle (KA) and hip-knee (HK) joint couplings were determined. There was a significant increase in variability of the KA and HK joint couplings in the flexion-extension axes during the loading and propulsion phases as hopping progressed. However, there was maintenance of performance output characteristics throughout the task. These findings suggest that changes in joint coupling variability may be a compensatory strategy to allow continuous single-leg hopping as the effects of muscular fatigue increase.

KEY WORDS: fatigue, joint coupling variability, motor control.

INTRODUCTION: Motor variability is inherent within human movement and may serve to optimise performance output and allow adaptation to changes in intrinsic and extrinsic factors such as fatigue or the environment (Hamill et al., 2012; Preatoni et al., 2013). Whether there is an “optimal” range of motor variability during repetitive tasks remains unknown. However, it has been suggested that large decreases in motor variability may lead to a rigid system while an increase in variability could yield an unstable system (Hamill et al., 2012; Preatoni et al., 2013). This may suggest that small increases in motor variability are an example of the normal modulation of movement strategies based on alterations to intrinsic and extrinsic factors, rather than a marker of maladaptation that may increase the risk of injury (Hamill et al., 2012). Fatigue is often induced during repetitive tasks, specifically during sporting and athletic activities. It has been demonstrated that fatigue during repetitive lower limb tasks leads to compensatory changes in ankle and knee function (Komi, 2000). These compensatory strategies may lead to subsequent alterations in multi-segment coordination and motor variability. However, the results from a number of studies investigating motor variability during fatiguing repetitive tasks have been conflicting. During a continuous run to exhaustion coordinative variability was shown to remain unchanged (Miller et al., 2008), whilst during treadmill walking fatigue of the tibialis posterior muscle (Ferber & Pohl, 2011) and the performance of a repetitive maximal vertical jump test (Dal Pupo et al., 2013) led to an increase in coordinative variability. The effect of fatigue on variability may be confounded by changes in the performance output of the task. Further, due to coordinative variability only being measured before and after the fatiguing protocol (Dal Pupo et al., 2013; Ferber & Pohl, 2011) and at the start, middle and end of the continuous task (Miller et al., 2008), task learning and subsequent change in motor patterning throughout the trial could confound the effect of fatigue. Thus, it is not clear whether lower limb coordinative variability is sensitive

to increases in fatigue during a repetitive task when performance output is controlled. Therefore, the purpose of this study was to examine the effect of fatigue on lower limb joint coupling variability (JCV) during repetitive lower limb loading when performance output remained unchanged.

METHODS: Twenty healthy recreationally active male and female participants completed a single-leg hopping trial to volitional exhaustion. Kinetic (AMTI, Gen 5, USA) and kinematic data (NDI, Optotrak, Canada) were collected synchronously at 1500 Hz and 150 Hz (First Principles software, Version 1.2.4), respectively. All trials were performed barefoot and on the participant's self-selected dominant leg. Consistent with previous research, a seven-segment model was used to model the trunk (G. Wu et al., 2005), pelvis, thigh, shank (Ball, 2011), hindfoot, forefoot (W. L. Wu et al., 2000) and hallux of the hopping leg. Following a warm-up and familiarisation trials, participants then completed a 15 s single-leg hopping trial from which each individual's target hop height was calculated from the sacral marker cluster. To control each individuals hop height, tactile feedback was provided via a custom built apparatus and set to the participants target hop height. Participants then completed one experimental trial to volitional exhaustion of on-the-spot, single-leg hopping at 2.2 Hz using an audible digital metronome and aiming to hop to the pre-determined target hop height. Volitional exhaustion was defined as the point when the participant could no longer maintain the required performance outcomes of hopping cadence and target hopping height.

Kinetic data were dual-pass filtered with a low pass Butterworth filter and 50 Hz cut-off. Kinematic data were interpolated using spline interpolation for up to a maximum gap of 10 frames and dual-pass filtered using a fourth order Butterworth filter with an 8 Hz cut-off (Visual 3D, Version 4). Derived variables included hopping frequency, vertical displacement of the centre of mass (COM) during flight (z_f) and loading phases (z_l) and JCV. Joint coupling variability was calculated for the knee flexion/extension-ankle flexion/extension (KxAx), hip flexion/extension-knee flexion/extension (HxKx) and knee flexion/extension-ankle eversion/inversion (KxAy) couplings during loading and propulsive phases. Joint angle data were normalised to 101 data points over loading and propulsive phases and coupling angles calculated as the orientation of the resultant vector to the right horizontal between two adjacent data points (Pollard et al., 2005). Calculated values were converted from radians to degrees and the standard deviation (SD) of the coupling angles across 10 hop cycles calculated on a point-by-point basis for the loading and propulsive phases, providing a measure of between-cycle, within-participant variability (Pollard et al., 2005). All derived variables were calculated as the mean of 10 consecutive hop cycles at time periods of 0, 20, 40, 60, 80 and 100% of the duration of each trial. A repeated measure ANOVA was used to determine differences and trends over time for all dependent variables. Post-hoc pairwise comparisons were made to further explore differences between each of the six hopping periods. Alpha levels were set *a priori* with significance accepted at $p < 0.05$.

RESULTS: Total hopping duration was (mean \pm SD) 75 \pm 34 s. There was no significant main effect detected between hopping frequency, z_f or z_l across the duration of the trial. There was a significant increase in KxAx, HxKx and KxAy during loading over time ($p < 0.001$) with a significant increasing linear trend ($p < 0.001$). Post hoc pairwise comparisons determined that KxAx and HxKx during loading were greater at 40 ($p = 0.015$; $p = 0.002$), 60 ($p = 0.014$; $p = 0.001$), 80 ($p = 0.005$; $p = 0.001$) and 100% ($p = 0.001$; $p = 0.001$) compared to the start of the trial (Figure 1). Post hoc pairwise comparisons for KxAy during loading revealed the 80 and 100% time periods were significantly greater than 0% ($p = 0.015$; $p = 0.001$) whilst 60% was only borderline significantly greater ($p = 0.042$) (Figure 1). There was a significant increase in KxAx and HxKx during propulsion over time ($p < 0.001$) with a significant

increasing linear trend ($p < 0.001$). Post hoc comparisons only revealed KxAx to significantly increase from 0 to 100% ($p = 0.004$) and HxKx to significantly increase from 0 to 60 ($p = 0.036$), 80 ($p = 0.037$) and 100% ($p = 0.002$) during propulsion (Figure 2).

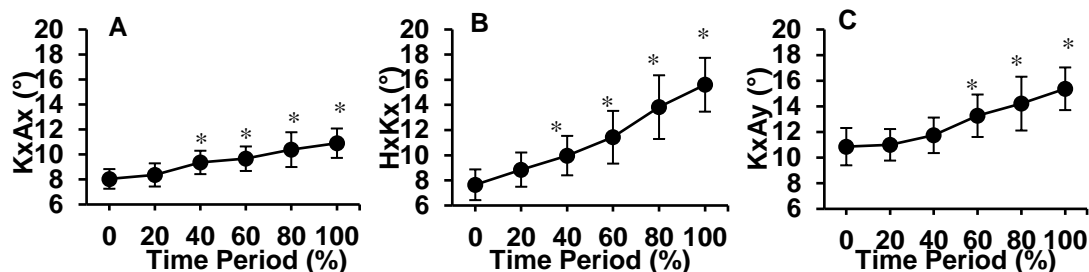


Figure 1: Mean (2SE) joint coupling variability (degrees) during loading at each time point for A) knee flexion/extension – ankle flexion/extension (KxAx); B) hip flexion/extension – knee flexion/extension (HxKx); and C) knee flexion/extension – ankle abduction/adduction (KxAy). * indicates $p < 0.05$ compared to 0% time period.

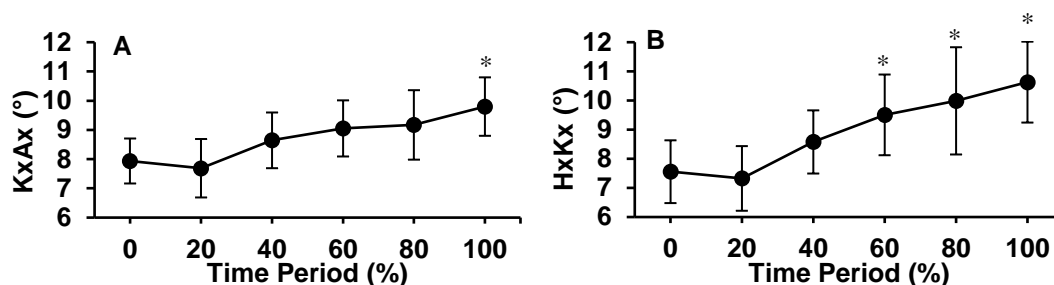


Figure 2: Mean (2SE) joint coupling variability (degrees) during propulsion at each time point for A) knee flexion/extension – ankle flexion/extension (KxAx); and B) hip flexion/extension – knee flexion/extension (HxKx). * indicates $p < 0.05$ compared to 0% time period.

DISCUSSION: This study demonstrated that there was greater JCV during both loading and propulsion phases as participants approached fatigue while motor performance remained unchanged. This finding supports previous research (Hamill et al., 2012; Preatoni et al., 2013) and suggests that motor variability is not detrimental to motor performance and in fact may allow the continuation of a repetitive submaximal task until volitional exhaustion. Increased JCV did not impact the performance with maintenance of spatiotemporal characteristics of the COM and may be appropriate and expected as a strategy to counter the effects of muscular or physiological fatigue (Dal Pupo et al., 2013; Singh et al., 2010). The preservation of performance output as fatigue increased may be due to co-variance between KxAx and HxKx during the loading and propulsion phases, demonstrated by a similar linear increase in the variability of both couplings. Co-variance between fatigued and less fatigued segments has been demonstrated in other repetitive tasks when the performance output was maintained such as with finger tapping (Singh et al., 2010), postural sway (Singh & Latash, 2011) and walking (Ferber & Pohl, 2011). However, these adaptive patterns had previously not been shown during a fatiguing and rapid loading task when performance outcomes were maintained. Co-variance has been suggested to be a mechanism to preserve the performance output of the fatiguing task by controlling the variability into a region of “good variance” (Singh et al., 2010). For example, if the variability of only the distal coupling was to increase without a subsequent increase in the variability of the proximal coupling the performance output of the task may be affected. Therefore, to maintain the performance output of the task

the less fatigued coupling co-varied with the fatigued coupling resulting in minimal impact on performance output.

Increased JCV may act as a preventative mechanism to minimise the risk of injury due to repetitive loading. Previous research has revealed greater lower limb variability in healthy compared to injured individuals (Hamill et al., 1999; Heiderscheit et al., 2002). They proposed that greater coordinative variability permitted healthy individuals numerous intra-segment motor patterns that may be used to successfully complete the continuous task. Therefore, increases in lower limb JCV during a fatiguing and repetitive task which does not affect the performance may allow the magnitude of forces to be attenuated between the coordinated limb segments.

CONCLUSION: Motor variability and co-variation at a kinematic level may be an essential strategy utilised by healthy individuals to prolong the detrimental effects of fatigue that can effect performance during a repetitive task (Singh et al., 2010). Further, increasing JCV may be a strategy used by healthy individuals to allow external forces to be distributed over numerous tissues, rather than repeatedly loading the same tissues. Therefore, increases in lower limb motor variability during a repetitive and rapid loading task which utilises the stretch-shorten cycle may serve to potentially lower the risk of lower limb overuse injuries as fatigue levels increase (Hamill et al., 2012; Singh & Latash, 2011).

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**Appendix E: Mudie et al., (2016). Conference abstract presented at
the 34th International Conference on Biomechanics in Sport**

**SEX DIFFERENCES IN LOWER LIMB MOVEMENT VARIABILITY DURING A
FATIGUING REPETITIVE LOADING TASK**

Kurt L. Mudie, Amitabh Gupta, Simon Green and Peter J. Clothier

**School of Science and Health, Western Sydney University,
Sydney, NSW, Australia**

This study evaluated differences in lower limb joint coupling variability between recreationally-active male ($n = 21$) and female participants ($n = 20$) during single-leg hopping to exhaustion. Spatio-temporal characteristics and variability of the knee-ankle and hip-knee joint couplings were determined over the duration of hopping. As fatigue progressed joint coupling variability increased by a greater magnitude in females compared to males. Females had significantly lower variability compared to males in the knee-ankle couplings during the propulsion phase at the beginning of the trial but this effect progressively disappeared during the trial. These findings suggest that as fatigue progresses, there is a regression to a similar magnitude of joint coupling variability which may represent a common level of synchronous joint interaction between sexes.

KEY WORDS: fatigue, joint coupling variability, motor control, hopping, stretch-shorten cycle.

INTRODUCTION: Variability within human movement is considered a normal and functional feature of the neuromuscular system (Preatoni et al., 2013). It is suggested that there is an optimal range of variability that permits flexibility to changes in intrinsic or extrinsic factors. Conversely, extreme levels of variability, i.e. too much or too little, may be associated with pathology (Hamill et al., 1999; Preatoni et al., 2013).

Interestingly, woman have demonstrated lower levels of movement variability than men during a side cut manoeuvre (Pollard et al., 2005) and treadmill running task (Barrett et al., 2008). It was suggested that the lower variability in woman may limit their ability to adapt to changes in task demands. Further, low levels of variability may expose the musculoskeletal system to repetitive loads. This may be problematic during fatiguing repetitive rapid loading tasks (Hamill et al., 1999).

Previous research has shown that fatigue increases movement variability during repetitive loading tasks (Dal Pupo et al., 2013; Ferber & Pohl, 2011; Mudie, Gupta, & Clothier, 2015). However, these studies only investigated males (Dal Pupo et al., 2013) or pooled males and females together (Ferber & Pohl, 2011; Mudie et al., 2015). Thus, it is not clear if the effect of fatigue on movement variability during a repetitive loading task differs between males and females. Therefore, the purpose of this study was to test the effect of sex on lower limb movement variability during a fatiguing repetitive loading task.

METHODS: Forty-one healthy recreationally active male ($n = 21$; 23 ± 3 years of age; 1.8 ± 0.1 m height; 76.4 ± 7.6 kg mass) and female ($n = 20$; 22 ± 3 years of age; 1.7 ± 0.1 m height; 61.3 ± 7.7 kg mass) participants completed a single-leg hopping trial to volitional exhaustion. Warm-up and familiarisation trials were completed barefoot and on the participant's dominant leg. Twenty-nine active and fifteen calibration markers were used to model the trunk, pelvis, thigh, shank, hindfoot, forefoot and hallux of the hopping leg (Mudie et al., 2015). To control task performance (hopping frequency and height), an audible metronome was set at 2.2 Hz and tactile feedback

set to each individuals target hop height. Target hop height was determined from a 15 s single-leg hopping trial at 2.2 Hz and calculated by a segmental analysis method. Participants then completed one single-leg hopping trial to volitional exhaustion, at 2.2 Hz to the pre-determined target hop height. Volitional exhaustion was defined as the point when the participant could no longer maintain the required performance outcomes of hopping frequency and target hopping height. Kinetic (AMTI, Gen 5, USA) and kinematic data (NDI, Optotrak, Canada) were recorded synchronously at 1500 Hz and 150 Hz (First Principles software, Version 1.2.4) respectively, and processed using Visual 3D (C-Motion, Version 4, USA).

Missing marker data were interpolated using spline interpolation for up to a maximum gap of 10 frames. Kinetic and kinematic data were filtered with a fourth order dual-pass Butterworth filter with a 50 Hz cut-off and a fourth order dual-pass Butterworth filter with an 8 Hz cut-off, respectively. A single hop cycle was defined as a complete flight phase and subsequent contact phase. The contact phase included a loading (initial contact to peak vertical ground reaction force) and propulsive (peak vertical ground reaction force to toe off) phase. Dependent variables for each hop cycle included hopping frequency, vertical displacement of the centre of mass (COM) during flight (z_i), vertical stiffness (K) and joint coupling variability. Vertical stiffness was calculated as the quotient of normalised force and COM displacement during the loading phase. Joint coupling variability was calculated for the knee flexion/extension-ankle flexion/extension (KxAx), hip flexion/extension-knee flexion/extension (HxKx) and knee flexion/extension-ankle eversion/inversion (KxAy) couplings during the loading and propulsive phases. Joint angle data were normalised to 101 data points over the loading and propulsive phases and coupling angles calculated as the orientation of the resultant vector to the right horizontal between two adjacent data points (Pollard et al., 2005) (Equation 1).

$$\Theta_i = \text{abs} [\tan^{-1} (y_{i+1} - y_i / x_{i+1} - x_i)] \quad (1)$$

where i = data point 1, 2 and n of the time series. Calculated values were converted from radians to degrees and the standard deviation (SD) of the coupling angle across 10 consecutive hop cycles was calculated on a point-by-point basis for the loading and propulsive phases, providing a measure of between-cycle, within-participant variability (Pollard et al., 2005). Each dependent variable was calculated as the mean of 10 consecutive hop cycles at time periods of 0, 20, 40, 60, 80 at 100% of the trial. A 6 (time period) x 2 (sex) repeated measure ANOVA was used to determine between sex differences over time for all dependent variables. If there was a significant interaction or main effect, post-hoc one-way ANOVA (time period) and independent samples t -tests (sex) were performed. Alpha levels were set *a priori* with significance accepted at $p < 0.05$. *Cohen's d* effect sizes were calculated and thresholds of 0.2 = small, 0.5 = moderate and 0.8 = large were used to qualitatively describe effect sizes.

RESULTS: Total hopping duration was (mean \pm SD) 79 \pm 30 s for males and 78 \pm 22 s for females ($p = 0.949$, $d = 0.056$). There were no significant effects of sex on performance output characteristics, including hopping frequency ($p = 0.843$), z_i ($p = 0.355$) and K ($p = 0.715$). However, there was a significant decrease in K for females only from the 0% time period to the 80% ($p = 0.029$, $d = -0.628$) and 100% ($p = 0.010$, $d = -0.588$) time periods.

During loading, there was a significant increase in KxAx for males from the 0% time period to the 80% ($p = 0.025$, $d = 1.026$) and 100% ($p = 0.007$, $d = 1.531$) time periods and females from the 0% time period to the 100% ($p = 0.002$, $d = 1.420$) time period (Figure 1). There was a significant increase in HxKx for males and females from the 0% time period to the 60% ($p = 0.002$, $d = 1.219$; $p = 0.003$, $d = 1.126$), 80% ($p = 0.001$, $d = 2.176$; $p < 0.001$, $d = 1.571$) and 100% ($p < 0.001$, $d = 2.946$; $p < 0.001$, $d = 2.962$) time periods and KxAy from the 0% time period to the 100% ($p = 0.013$, $d =$

1.424; $p = 0.001$, $d = 1.906$) time period (Figure 1). In contrast, there was no significant difference between males and females for KxAx ($p = 0.063$), HxKx ($p = 0.545$) and KxAy ($p = 0.133$) (Figure 1). However, females tended to have lower variability than males with a moderate effect size in KxAx at the 80% ($d = 0.605$) time period and in KxAy at the 0% ($d = 0.621$) time period (Figure 1).

During propulsion, there was a significant increase in KxAx for females only from the 0% time period to the 100% ($p = 0.017$, $d = 0.971$) time period (Figure 2). There was a significant increase in HxKx for females from the 0% time period to the 80% ($p = 0.007$, $d = 1.059$) and 100% ($p < 0.001$, $d = 1.838$) time periods and males from the 0% time period to the 100% ($p = 0.028$, $d = 1.292$) time period (Figure 2). Females were significantly lower than males in KxAx at the 40% ($p = 0.033$, $d = 0.591$) and 60% ($p = 0.043$, $d = 0.668$) time periods and in KxAy at the 20% ($p = 0.038$, $d = 0.703$) time period (Figure 2). Females tended to have lower variability than males with a moderate effect size in KxAx at the 0% ($p = 0.068$, $d = 0.640$) and 20% ($p = 0.064$, $d = 0.626$) time periods and in KxAy at the 0% ($p = 0.071$, $d = 0.628$) time period (Figure 2).

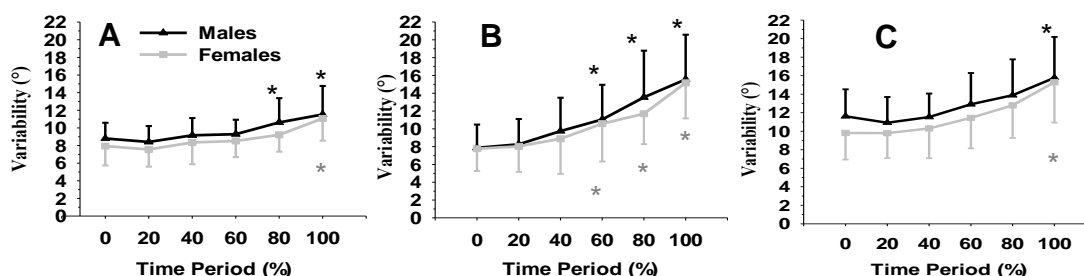


Figure 1: Mean (SD) joint coupling variability ($^{\circ}$) during loading for males (black) and females (grey) at each time period. A) Knee flexion/extension–ankle flexion/extension (KxAx); B) hip flexion/extension–knee flexion/extension (HxKx); and C) knee flexion/extension–ankle abduction/adduction (KxAy). * = $p < 0.05$ compared to 0% time period for males (black) and females (grey).

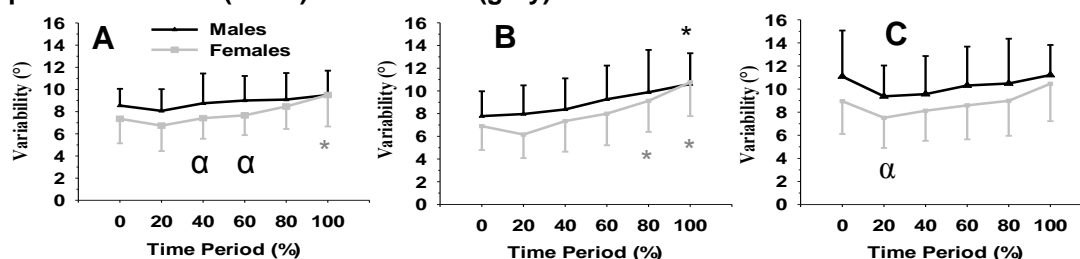


Figure 2: Mean (SD) joint coupling variability ($^{\circ}$) during propulsion for males (black) and females (grey) at each time period. A) Knee flexion/extension–ankle flexion/extension (KxAx); B) hip flexion/extension–knee flexion/extension (HxKx); and C) knee flexion/extension–ankle abduction/adduction (KxAy). * = $p < 0.05$ compared to 0% time period for males (black) and females (grey). α = $p < 0.05$ between males and females at the specific time period.

DISCUSSION: This study demonstrated males and females increased joint coupling variability as fatigue progressed during a repetitive loading task. Further, females had lower coupling variability compared to males during the first half of the task. As participants approached exhaustion the difference between the sexes gradually decreased as females increased joint coupling variability by a greater magnitude than males. This suggests that during the continued performance of a repetitive loading task, males and females regress to similar patterns of joint coupling variability.

These results are in agreement with previous studies (Barrett et al., 2008; Pollard et al., 2005) demonstrating females to have lower variability than males during a non-fatigued side cut and running task. A key extension of previous research was that the

current task was performed to exhaustion. Further, there were no differences in performance output characteristics between the sexes over the duration of the trial. Although performance output was similar between sexes during the trial, as fatigue progressed females increased their joint coupling variability by a greater magnitude than the males. Resulting in the sexes regressing to similar levels of joint coupling variability as fatigue progressed. Previous research has suggested that lower joint coupling variability may lead to a more localised loading of lower limb musculature (Hamill et al., 1999; Heiderscheit et al., 2002). The finding that females had lower joint coupling variability during the first half of the trial may potentially risk females to a greater exposure of a number of cycles with similar loading to tissues within the kinetic chain compared to males. If low coupling variability is considered detrimental to the musculoskeletal system, it is possible females may aim to adapt variability levels to a more 'optimal' range as the task progresses. Therefore, although females may inherently possess lower coupling variability than males, they have the capacity to achieve similar values under particular conditions.

Significant differences between the sexes in joint coupling variability were only shown during the propulsive phase and were smaller in magnitude compared to previous research which had evaluated a side-cut task (Pollard et al., 2005). It is possible both the lower magnitude of difference between sexes and lower absolute values of joint coupling variability in this study compared to Pollard et al., (2005), may be due to being task specific. A single-leg hopping task to a consistent target height and frequency is a highly controlled task compared to other tasks that have been used. On-the-spot single-leg hopping has less available degrees of freedom than other more complex tasks such as running or a side-cut, potentially limiting the amount of variability available at a joint level. The current results demonstrate only subtle differences in joint coupling variability between the sexes during a simple, controlled repetitive task performed to exhaustion. Therefore, differences in variability between men and woman may be relative to the type of task and it is plausible that during a more complex task or within an unpredictable environment, there may be greater differences in variability between the sexes.

CONCLUSION: Although females have lower joint coupling variability than males at the start of a repetitive loading task, as fatigue progresses females increase joint coupling variability by a greater magnitude than males. This results in the sexes regressing to similar levels of joint coupling variability during a fatiguing repetitive loading task. Therefore, although females may inherently possess lower joint coupling variability than males, the capacity to achieve similar values to males under particular conditions has been demonstrated.

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Appendix F: Letter of ethical approval

Locked Bag 1797
Penrith NSW 2751 Australia
Office of Research Services
ORS Reference: H10704



HUMAN RESEARCH ETHICS COMMITTEE

14 July 2014

Doctor Peter Clothier

School of Science and Health

Dear Peter,

I wish to formally advise you that the Human Research Ethics Committee has approved your research proposal H10704 "An evaluation of motor control in repeated submaximal loading tasks.", until 1 May 2017 with the provision of a progress report annually if over 12 months and a final report on completion.

Conditions of Approval

1. A progress report will be due annually on the anniversary of your approval date.
2. A final report will be due at the expiration of your approval period as detailed in the approval letter.
3. Any amendments to the project must be approved by the Human Research Ethics Committee prior to the project continuing. Amendments must be requested using the HREC Amendment Request Form:
http://www.uws.edu.au/_data/assets/pdf_file/0018/491130/HREC_Amendment_Request_Form.pdf
4. Any serious or unexpected adverse events on participants must be reported to the Human Ethics Committee as a matter of priority.
5. Any unforeseen events that might affect continued ethical acceptability of the project should also be reported to the Committee as a matter of priority
6. Consent forms are to be retained within the archives of the School or Research Institute and made available to the Committee upon request

Please quote the registration number and title as indicated above in the subject line on all future correspondence related to this project. All correspondence should be sent to the email address humanethics@uws.edu.au.

This protocol covers the following researchers:

Peter Clothier, Kurt Mudie, Amitabh Gupta

Yours sincerely

A handwritten signature in black ink, appearing to read "E Deane".

Professor Elizabeth Deane
Presiding Member,
Human Researcher Ethics Committee

Appendix G: Information for participants



School of Science and Health

Motor Control during Repeated Loading

Information for Participants

Dear potential research participant,

Our names are Dr Peter Clothier, Dr Amitabh Gupta and Mr Kurt Mudie and we would like to extend an invitation for you to participate in a research project we are conducting titled: *“An evaluation of motor control in running, jumping and hopping tasks”*. We have attached below information detailing what the project involves and your requirements as a participant. If you are interested in participating, please contact us on the details provided in the information document and we will send you a Pre-Exercise Health Screening questionnaire to assess your suitability as a potential research participant.

If you have any further questions regarding this project, please don't hesitate to contact any of us. Thank you for considering this invitation.

Thank you for your consideration,

Dr Peter Clothier, Dr Amitabh Gupta and Mr Kurt Mudie

(Principal Researchers, University of Western Sydney)

Researchers:

- Dr Peter Clothier, University of Western Sydney (Ph: 02 4620 3743; Email: P.Clothier@uws.edu.au)

- Dr Amitabh Gupta, University of Western Sydney (Ph: 02 4620 3757; Email: A.Gupta@uws.edu.au)

- Mr Kurt Mudie, University of Western Sydney (Ph: 0405 259 557; Email: K.Mudie@uws.edu.au)

Project title

“An evaluation of motor control in running, jumping and hopping tasks ”.

Who is carrying out the study?

You are invited to participate in a study conducted by the following listed researchers from the School of Science and Health, the University of Western Sydney.

- Dr Peter Clothier (Principal Researcher)
University of Western Sydney – School of Science & Health
Ph: 02 4620 3743 Email: p.clothier@uws.edu.au

- Dr Amitabh Gupta (Principal Researcher)
University of Western Sydney – School of Science & Health
Ph: 02 4620 3757 Email: a.gupta@uws.edu.au

- Mr Kurt Mudie (Principal Researcher)
University of Western Sydney – School of Science & Health
Ph: 0405 259 557 Email: k.mudie@uws.edu.au

What is the study about?

This document provides information to potential participants for a research project investigating motor control adaptations during running, jumping and hopping tasks.

Details regarding the specific requirements of participants, potential risks and benefits and other information are contained below.

Note: To be eligible to participate in this research you must:

- Pass a pre-exercise health screening questionnaire.
- Be free from injury which may adversely affect your performance or ability to complete required tasks.
- Currently be exercising or participating in an activity involving running for between 1 - 4 hours a week and have been doing so for a minimum of 6 months, OR training for your chosen sport consistently between 10 – 20 hours a week for at least 3 years prior to testing.
- Report no past or current history of lower limb pathology, injury or pain, lower limb fracture or surgery OR report a past history of soft tissue injury (eg. Achilles or patellar tendinopathy; ankle sprain; quadriceps/hamstring/calf muscle strain; anterior cruciate ligament injury managed either non-surgically or surgically; medial tibial stress syndrome) that does not restrict the amount or intensity of your current activity and report no past history of lower limb or spinal fracture, ongoing lower limb pathology, pain or injury and systemic illness or disease.
- Be aged between of 18 - 42 years of age.

What does the study involve?

Experimental testing will take place at the Building 24 Human Movement Laboratory at the University of Western Sydney – Campbelltown Campus.

As a participant you will be required to wear above knee length sporting shorts and a shirt which will not impede the palpation of bony landmarks. All hopping and running trials are to be performed barefoot. Small electrodes and markers will be adhered to the surface of the lower limb and trunk to record muscle activity levels and lower limb movement during the hopping, running and drop jump trials.

A 5 minute warm up on a treadmill will be performed prior to testing at a walking/light jogging pace (6 – 8 km/h). Once the warm up is complete you will perform a series of familiarisation trials to ensure you understand the testing protocol and can match your hopping and running to specific task constraints, i.e. hopping frequency and running speed.

The experimental testing protocol may require you to perform trials of on-the-spot hopping and landing during drop jumps on a force platform and steady state running on a treadmill. Hopping will be performed at a specified frequency of 2.2 Hz at a predetermined set height to volitional exhaustion (cannot hop anymore due to fatigue) and running performed on a treadmill for a set duration at a steady state of 10 km/h (light jog). Drop jumps will be performed from a pre-determined set height during both exhausted and non-exhausted states. A metronome will provide the hopping frequency which will control the rate at which you will hop by generating an audible sound at a pre-determined frequency (2.2 Hz). Hop height will be controlled via tactile feedback (head touching a soft height marker), running speed via the treadmill and drop jumps by an adjustable bar set at a predetermined height. Appropriate rest periods will be employed between each condition. Once all trials have been completed an appropriate cool down will be prescribed. Trials will cease immediately if at any time you are feeling ill, are uncomfortable, sustain an injury or voluntarily decide not to continue participating.

How much time will the study take?

As a participant you will be required for approximately 120 minutes. This includes the time required to complete the informed consent form, preparation, familiarisation and the experimental trials.

Will the study benefit me?

The study aims to further biomechanical knowledge of human performance with results potentially influencing future research questions and directions. No direct benefits to you are expected from participation. However, you may receive an educational benefit from being exposed to scientific experimental research process. This would be particularly so if you are studying an Exercise and Sport Science degree or a unit in biomechanics.

The risk associated with the experimental protocol for this research is considered low. The exercise trials will involve a low to moderate paced treadmill running task, volitional exhaustion single/double-leg hopping trials and drop jumps from a predetermined set height. During these activities you may experience a degree of muscular fatigue and perspiration. You will also experience other normal responses to exercise which include increased heart rate and increased blood pressure. Mild discomfort may be experienced when removing the electrodes and markers (similar to removal of a band-aid).

However, the level of risk is no greater than what can be expected during low to moderate intensity exercise. Therefore, if you meet the eligibility criteria for this research you are considered to be at low risk of injury as you will be regularly exercising at or beyond the intensity required during the test trials.

Nonetheless, every effort will be made to minimise any associated participant risk by:

- Evaluation of preliminary information relating to your health & fitness;
- Conducting all trials indoors to minimise environmental effects (heat/cold);
- Engaging you in an appropriate warm-up prior to testing to minimise risk of injury;
- Observation of your technique during the warm-up and experimental trials.

In the unlikely event that injury does occur during the testing, the attending researchers will initiate an appropriate first aid treatment and action plan.

How is the study being paid for?

This research project is funded by the University of Western Sydney's School of Science and Health. Your involvement in this project will be on the basis of an unpaid volunteer, participation will be at no additional cost to you and the researchers will try to ensure the experimental testing will cause minimal inconvenience to you.

Will anyone else know the results? How will the results be disseminated?

All aspects of the study, including results, will be confidential and only the researchers will have access to information on participants except as required by law. As a participant, all information collected from you will be coded by numbers and lettering and stored separately from any listing that includes your name to ensure the maintenance of privacy and confidentiality. Throughout this investigation all data will be securely maintained in the possession of Dr Peter Clothier (Principal Researcher), Dr Amitabh Gupta (Principal Researcher) and Mr Kurt Mudie (Principal Researcher). Following completion of the study, all information collected for, used in or generated by this project will be disposed of after 5 years.

Furthermore, your personal details and data will not be released or revealed to any other party without your written consent. The de-identified and aggregated results from this investigation are intended to be used for publication in a scientific domain (presentation and journal publication). You are assured that this will be achieved whilst maintaining your right to privacy and confidentiality.

Can I withdraw from the study?

Participation is entirely voluntary. You are not obliged to participate and if you do participate you can withdraw at any time without giving any reason. Whatever your decision there will be no consequences and it will not affect your relationship with the researchers in any way. This includes any perceived impact between or on the student - researcher (lecturer) relationship.

Can I tell other people about the study?

Yes, you can tell other people about the study and provide them with researcher contact details. They are then welcome to contact one of the researchers to discuss their potential participation in the research project and obtain the information sheet.

What if I require further information?

When you have read this information, Mr Kurt Mudie will discuss it with you further and answer any questions you may have. If you would like to know more at any stage please feel free to contact:

- Mr Kurt Mudie (Principal Researcher)
University of Western Sydney – School of Science & Health

Ph: 0405 259 557 Email: k.mudie@uws.edu.au

- Dr Peter Clothier (Principal Researcher)
University of Western Sydney – School of Science & Health

Ph: 02 4620 3743 Email: p.clothier@uws.edu.au

- Dr Amitabh Gupta (Principal Researcher)
University of Western Sydney – School of Science & Health

Ph: 02 4620 3757 Email: a.gupta@uws.edu.au

What if I have a complaint?

This study has been approved by the University of Western Sydney Human Research Ethics Committee. The Approval Number is H10704.

If you have any complaints or reservations about the ethical conduct of this research, you may contact the Ethics Committee through the Office of Research Services (Ph: +61 2 4736 0229 or Fax +61 2 4736 0013; Email: humanethics@uws.edu.au).

Any issues you raise will be treated in confidence and investigated fully, and you will be informed of the outcome. If you agree to participate in this study, you may be asked to sign the Participant Consent Form.

The information sheet is for the participant to keep and the consent form is retained by the researcher.

Appendix H: Pre-exercise health screen questionnaire



School of Science and Health

Motor Control during Repeated Loading

Pre-Exercise Health Screening

The information provided in the answers of this health screen is required by the researchers to assess your suitability for participation in the research project titled "*An evaluation of motor control in repeated submaximal loading tasks*". If you do not understand a question please do not hesitate to contact either of the researchers who are listed on the information for participation sheet.

PLEASE ANSWER ALL QUESTIONS HONESTLY

Date of Birth: / / **Sex:** Male / Female

Researchers Only

Subject code: _____

Group:

Healthy Untrained	<input type="checkbox"/>	Injured Untrained	<input type="checkbox"/>
Healthy Trained	<input type="checkbox"/>	Injured Trained	<input type="checkbox"/>

Section 1 – Signs and Symptoms

1. Has your doctor ever told you that you have a heart condition or have you ever suffered a stroke? (Please Circle)

No Yes

2. Do you ever experience unexplained pains in your chest at rest or during physical activity/exercise?

No Yes

3. Do you ever feel faint or have spells of dizziness during physical activity/exercise that causes you to lose balance?

No Yes

4. Have you had an asthma attack requiring immediate medical attention at any time over the last 12 months?

No Yes

5. If you have diabetes (type I or type II) have you had trouble controlling your blood glucose in the last 3 months?

No Yes

6. Do you have any diagnosed muscle, bone or joint problems that you have been told could be made worse by participating in physical activity/exercise?

No Yes

If yes, provide details. If no, leave blank.

7. Do you have any other medical condition(s) that may make it dangerous for you to participate in physical activity/exercise?

No Yes

If yes, provide details. If no, leave blank.

Section 2 – Risk Factors

1. Do you have a family history of heart disease (eg: stroke, heart attack)? That is has your father, mother, brother, sister had a stroke/heart attack?

No Yes

If you answered YES to the previous question, what is/are the gender(s) and age(s) of the relative(s) with heart disease?

Male > 55 Male < 55

Female > 65

Female < 65

2. Do you smoke cigarettes on a daily or weekly basis or have you quit smoking in the last 6 months?

No

Yes

If you answered YES to the previous question, how many cigarettes do you currently or did smoke per day / week?

Please provide details below.

3. Which statement best describes your current physical activity / exercise level?

Sedentary

Light

Moderate

Vigorous

4. How many exercise / physical activity sessions do you usually do each week?

5. How many minutes of exercise / physical activity do you usually do each week?

6. How long have you been continuously participating in the above activity?

_____ years _____ months _____ weeks

Have you been told that you have high blood pressure?

No Yes

7. Have you been told that you have high cholesterol?

No Yes

8. Have you been told that you have high blood sugar?

No Yes

9. Have you spent time in hospital (including day admission) for any medical condition/illness/injury during the last 12 months?

No Yes

If yes, please provide details below. If no, leave blank.

10. Are you currently taking a prescribed medication(s) for any medical conditions(s)?

No Yes

If yes, please list below the medical condition(s) and medication(s). If no, leave blank.

11. Are you pregnant or have you given birth within the last 12 months?

No Yes

If yes, please provide details below. If no, leave blank.

12. Do you have any muscle, bone or joint pain or soreness that is made worse by particular types of activity?

No Yes

If yes, please provide details below. If no, leave blank.

13. Have you previously sustained any leg injury (sprains, broken bones, muscle damage etc.)?

No Yes

If yes, please provide details below. If no, leave blank.

14. Does this injury still cause pain or influence your performance during
exercise/sport? No Yes

If yes, please provide details below. If no, leave blank.

15. Do you currently have any injury or illness that you believe may exclude you
from participation in this study?

No Yes

If yes, please give details

16. I believe that to the best of my knowledge, all of the information I have supplied
within this survey is correct.

No Yes

Declaration:

I, _____ have completed all questions honestly and to the best of my knowledge. I will inform one of the researchers if there are any changes in my health or injury status which may impact on my ability to participate in the study "The effect of augmented feedback on biomechanical and neuromuscular activity in the lower limb during single-leg hopping to volitional fatigue". An evaluation of the effect of graded compression garments on dynamic tasks."

Participant signature: _____

Participant name: _____

Date: _____

Appendix I: Informed consent form

School of Science and Health

*Motor Control during Repeated Loading***Informed Consent Form****Project title**

An evaluation of motor control during running, jumping and hopping tasks.

Researchers

Dr Peter Clothier	University of Western Sydney	ph: 02 4620 3743
Dr Amitabh Gupta	University of Western Sydney	ph: 02 4620 3757
Mr Kurt Mudie	University of Western Sydney	ph: 0405 259 557

If you agree to participate in this study, please sign both copies of this informed consent form, retain one copy for your records and return the other copy to the researchers.

Informed Consent Agreement

As a prospective participant in the above titled research project,

I, _____ declare that:

- The research program in which I am volunteering to participate has been explained fully to me both verbally and in writing.
- I have read and understood the accompanying information sheet, understand what my involvement will be and have had any questions regarding the research and my involvement answered to my satisfaction.
- I understand that the experimental testing protocol may require me to perform trials of on-the-spot hopping and landing during drop jumps on a force platform and running on a treadmill.

- I understand that the performance of the same or different tasks may be assessed before and after an intervention. Hence, there may be a requirement for some participants to attend more than one session in this project.
- I understand I am free to withdraw my consent at any time during the study, without explanation and without penalty, in which event my participation in the research project will immediately cease and any information obtained from it will not be used.
- I understand that whether or not I decide to participate, my decision will not affect my relationship with the researchers or the University of Western Sydney.
- I understand that all information I provide, including questionnaires will be coded by number and lettering and stored separately from any listing that includes my name and contact details.
- I understand that aggregated results from this research will be used for research purposes and may be reported in scientific/academic journals.
- I understand that the publication of data from this research will not individually identify me in any way.
- **I consent to participate as a volunteer in this research project.**

Signature of Participant: _____

Name of Participant: _____

Date: _____

This study has been approved by the University of Western Sydney Human Research Ethics Committee. The Approval number is: H1074

If you have any complaints or reservations about the ethical conduct of this research, you may contact the Ethics Committee through the Office of Research Services on Tel +61 2 4736 0229 Fax +61 2 4736 0013 or email humanethics@uws.edu.au . Any issues you raise will be treated in confidence and investigated fully, and you will be informed of the outcome.

Appendix J: Marker locations

Segment	Marker Placement		
	Marker Cluster	Calibration Markers	Active Markers
Thorax/ Abdomen	Upper back, overlying T8	Spinous process of C7 Spinous process of T8 Suprasternal notch Xiphoid process Left acromioclavicular joint Right acromioclavicular joint	
Pelvis	Sacrum	Left anterior superior iliac spine Right anterior superior iliac spine Left posterior superior iliac spine Right posterior superior iliac spine	
Thigh	Distal lateral end	Greater trochanter Medial epicondyle Lateral epicondyle	
Shank	Distal lateral end	Medial malleolus Lateral malleolus	
Hindfoot		Midpoint between medial and lateral calcaneal markers ^a	Posterior calcaneus inferior
Forefoot		53% of distance between heads of the first and fifth metatarsals ^a Midpoint between bases of the first and fifth metatarsals ^a	Sustentaculum tali Lateral calcaneus Base first metatarsal Base fifth metatarsal
Toe			Head first metatarsal Head fifth metatarsal Medial proximal phalanx

Note: ^a – Calibration markers were used to define segment coordinate systems and axes created in Visual 3D.

Appendix K: Local segment coordinate systems

Table K.1 Definition of the local coordinate system for the trunk/thorax, thigh, shank, hindfoot, forefoot and hallux of the participant's dominant leg and thus hopping leg (Grood & Suntay, 1983; G. Wu et al., 2002; G. Wu et al., 2005).

Local Coordinate System	Description
Trunk/thorax	
Origin	Suprasternal notch
Superior-inferior (z)	The line that connects the midpoint between xiphoid process and T8 and the midpoint between suprasternal notch and C7 pointing superiorly.
Medio-lateral (x)	The line perpendicular to the plane formed by suprasternal notch, C7 and the midpoint between xiphoid process and T8, pointing to the right.
Anterior-posterior (y)	The common line perpendicular to the Z and X axis, pointing forward.
Pelvis	
Origin	Right or left hip joint centre
Medio-lateral (x)	The line parallel to a line connecting the right and left ASIS's and pointing to the right
Anterior-posterior (y)	The line parallel to a line lying in the plane defined by the two ASIS's and the midpoint of the two PSIS's, orthogonal to the Z-axis and pointing anteriorly
Superior-inferior (z)	The line perpendicular to both medio-lateral and anterior-posterior, pointing cranially
Thigh	
Origin	Right or left hip centre, coincided with that of the pelvic origin
Superior-inferior (z)	The line joining the midpoint between the medial and lateral Femoral epicondyles and the origin, pointing cranially
Medio-lateral (x)	The line perpendicular to the superior-inferior axis, lying in the plane defined by the origin and the two femoral epicondyles, pointing to the right
Anterior-posterior (y)	The line perpendicular to both superior-inferior and medio-lateral axis, pointing anteriorly

Shank	
Origin	Coincident with midpoint between medial and lateral malleolus
Medio-lateral (x)	The line connecting medial and lateral malleolus and pointing to the right
Anterior-posterior (y)	The line perpendicular to the torsional plane of the tibia/fibula (plane defined by the vertical axis and the vector from the medial and lateral malleoli) and pointing anteriorly
Superior-inferior (z)	The common line perpendicular to the anterior-posterior and medio-lateral axis, pointing cranially
Hindfoot	
Origin	Coincident with that of the shank being midpoint between medial and lateral malleolus
Superior-inferior (z)	The line coincident with the long axis of the shank pointing cranially
Anterior-posterior (y)	The line perpendicular to the frontal plane of the shank (through the malleoli) and parallel to the floor, pointing anteriorly
Medio-lateral (x)	The common line perpendicular to the anterior-posterior and superior-inferior axis pointing to the right
Forefoot	
Origin	Midpoint between the base of the first and fifth metatarsals
Superior-inferior (z)	Perpendicular to the transverse plane containing the head of the first metatarsal and the base and head of the fifth metatarsal, pointing superiorly
Anterior-posterior (y)	Projected onto the transverse plane in a line from the midpoint of the base of the first and fifth metatarsals to the TOE marker (53% of the distance between the heads of the first and fifth metatarsals) anteriorly
Medio-lateral (x)	Mutually perpendicular to the superior-inferior and anterior-posterior axis, to the right
Hallux	
Origin	Head of the first metatarsal
Medio-lateral (x)	Aligned with lateral axis of forefoot, pointing to the right
Anterior-posterior (y)	From head of the first metatarsal (same height of hallux) to hallux, pointing anteriorly

Appendix L: Statistical results for Chapter 4

Table L.1 Two-tailed paired sample *t*-test results (degrees of freedom (*df*), *t* value and *p* value) for hop height compared PRE and POST trials

	<i>df</i>	<i>t</i>	<i>p</i>
Hop Height	37	7.610	0.001*

* = $p < 0.05$

Table L.2 One-way ANOVA results (degrees of freedom (*df*), *F* value, *p* value and observed power ($1-\beta$)) for the comparison of the performance output characteristics over the duration of the exhaustive hopping trial

	<i>df</i>	<i>F</i>	<i>p</i>	$1-\beta$
Hopping Frequency	3.342	2.066	0.101	0.550
Hopping Height	2.677	1.607	0.197	0.388
Vertical Stiffness	3.126	4.782	0.003*	0.903
Contact Time	2.800	10.592	0.001*	0.998

* = $p < 0.05$

Table L.3 One-way ANOVA results (degrees of freedom (*df*), *F* value, *p* value and observed power ($1-\beta$)) for the comparison of the joint coupling variability of the knee flexion/extension - ankle flexion/extension (KxAx), hip flexion/extension - knee flexion/extension (HxKx) and knee flexion/extension - ankle eversion/inversion (KxAy) couplings during the loading phase over the duration of the exhaustive hopping trial

	<i>df</i>	<i>F</i>	<i>p</i>	$1-\beta$
KxAx	3.251	20.198	0.001*	1.000
HxKx	2.986	48.389	0.001*	1.000
KxAy	3.204	23.312	0.001*	1.000

* = $p < 0.05$

Table L.4 One-way ANOVA results (degrees of freedom (*df*), *F* value, *p* value and observed power ($1-\beta$)) for the comparison of the joint coupling variability of the knee flexion/extension - ankle flexion/extension (KxAx), hip flexion/extension - knee flexion/extension (HxKx) and knee flexion/extension - ankle eversion/inversion (KxAy) couplings during the propulsion phase over the duration of the exhaustive hopping trial

	<i>df</i>	<i>F</i>	<i>p</i>	$1-\beta$
KxAx	3.772	8.497	0.001*	0.998
HxKx	3.595	25.259	0.001*	1.000
KxAy	3.454	5.129	0.001*	0.942

* = $p < 0.05$

Appendix M: Surface EMG electrode locations

Table M.1 Surface electromyography electrode locations and isometric test to confirm muscle location and test electromyography signal

Muscle	Electrode Location	Test
Gluteus medius	50% on the line from the iliac crest to the greater trochanter	Hip abduction
Vastus lateralis	2/3 on the line from the ASIS to the lateral boarder of the patella	Knee extension without rotating the thigh
Biceps femoris	50% on the line between the ischial tuberosity and the lateral epicondyle of the tibia	Knee flexion
Medial gastrocnemius	The most prominent bulge of the muscle	Plantar flexion of the foot in a standing position
Soleus	2/3 of the line between the medial condyle of the femur to the medial malleolus	In a flexed knee position (crouched) the participant lifted the heel from the floor
Tibialis anterior	1/3 on the line between the head of the fibula and the tip of the medial malleolus	Ankle dorsiflexion without extension of the great toe
Fibularis longus	25% on the line between the head of the fibula to the tip of the lateral malleolus	Eversion of the foot with plantar flexion of the ankle joint

Appendix N: Statistical results for Chapter 5

Table N.1 One-way ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) for the comparison of the duration of the feedforward phase of the gluteus medius (GM), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL) and fibularis longus (FL) muscles over the duration of the exhaustive hopping trial

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM	5.00	3.15	0.009*	0.87
VL	3.92	2.98	0.022*	0.85
MG	3.15	43.70	0.001*	1.00
SOL	3.44	10.17	0.001*	1.00
FL	3.48	4.70	0.002*	0.92

* = $p < 0.05$

Table N.2 One-way ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) for the comparison of the total duration of gluteus medius (GM), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscle activity over the duration of the exhaustive hopping trial

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM	3.96	2.81	0.028*	0.75
VL	3.61	11.30	0.000*	1.00
MG	3.30	12.16	0.001*	1.00
SOL	3.32	9.44	0.001*	1.00
FL	5.00	5.62	0.001*	0.98
TA	5.00	5.32	0.001*	0.99

* = $p < 0.05$

Table N.3 One-way ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) for the comparison of the MAV of the gluteus medius (GM), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL) and fibularis longus (FL) muscles during the feedforward phase over the duration of the exhaustive hopping trial

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM	3.24	0.00	1.000	0.05
VL	2.27	17.95	0.001*	1.00
MG	2.32	5.11	0.006*	0.85
SOL	2.30	37.28	0.001*	1.00
FL	2.51	8.27	0.001*	0.98

* = $p < 0.05$

Table N.4 One-way ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) for the comparison of the MAV of the gluteus medius (GM), biceps femoris (BF), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscles during the background phase over the duration of the exhaustive hopping trial

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM	3.53	3.25	0.018*	0.79
BF	2.33	3.35	0.033*	0.67
VL	2.04	9.72	0.001*	0.98
MG	2.64	1.88	0.145	0.44
SOL	2.30	22.38	0.001*	1.00
FL	2.21	3.11	0.045*	0.61
TA	1.92	5.51	0.007*	0.83

* = $p < 0.05$

Table N.5 One-way ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) for the comparison of the MAV of the gluteus medius (GM), biceps femoris (BF), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscles during the M1 phase over the duration of the exhaustive hopping trial

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM	3.21	5.11	0.002*	0.93
BF	3.70	8.99	0.001*	1.00
VL	2.20	16.65	0.001*	1.00
MG	3.21	5.87	0.001*	0.96
SOL	2.33	38.45	0.001*	1.00
FL	2.72	4.18	0.010*	0.82
TA	2.30	0.00	1.000	0.05

* = $p < 0.05$

Table N.6 One-way ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) for the comparison of the MAV of the gluteus medius (GM), biceps femoris (BF), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscles during the M2 phase over the duration of the exhaustive hopping trial

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM	2.04	10.23	0.001*	0.99
BF	3.01	9.28	0.001*	1.00
VL	2.45	12.97	0.001*	1.00
MG	2.58	5.43	0.003*	0.90
SOL	2.58	22.57	0.001*	1.00
FL	2.47	3.13	0.038*	0.65
TA	3.05	2.25	0.085	0.56

* = $p < 0.05$

Table N.7 One-way ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) for the comparison of the MAV of the gluteus medius (GM), biceps femoris (BF), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscles during the M3 phase over the duration of the exhaustive hopping trial

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM	1.97	8.91	0.001*	9.67
BF	3.36	9.61	0.001*	1.00
VL	1.99	15.15	0.001*	1.00
MG	3.54	1.01	0.401	0.29
SOL	2.34	10.09	0.001*	0.99
FL	3.12	3.59	0.015*	0.79
TA	2.81	4.87	0.004*	0.88

* = $p < 0.05$

Table N.8 One-way ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) for the comparison of the MAV of the gluteus medius (GM), biceps femoris (BF), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscles during the propulsion phase over the duration of the exhaustive hopping trial

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM	2.53	9.56	0.001*	0.99
BF	2.36	0.00	1.000	0.05
VL	1.95	15.12	0.001*	1.00
MG	2.79	8.62	0.001*	0.99
SOL	2.54	6.34	0.001*	0.94
FL	2.81	0.00	1.000	0.05
TA	3.10	19.41	0.001*	1.00

* = $p < 0.05$

Table N.9 One-way ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) for the comparison of the co-activation of the vastus lateralis and biceps femoris (VL:BF) muscle pair and the soleus and tibialis anterior (SOL:TA) muscle pair during each time phase over the duration of the exhaustive hopping trial

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
Background				
VL:BF	2.13	0.73	0.495	0.173
SOL:TA	2.31	9.41	0.001*	0.99
M1				
VL:BF	3.15	2.07	0.104	0.53
SOL:TA	2.53	9.43	0.001*	0.99
M2				
VL:BF	2.11	1.45	0.240	0.31
SOL:TA	3.00	3.34	0.022*	0.75
M3				
VL:BF	3.26	2.14	0.094	0.56
SOL:TA	2.53	1.52	0.219	0.36

* = $p < 0.05$

Appendix O: Statistical results for Chapter 6

Table O.1 Two-tailed independent sample *t*-test results (degrees of freedom (df), *t* value and *p* value) for total hopping duration compared between males and females

Total Duration	<i>df</i>	<i>t</i>	<i>p</i>
Male * Females	39	0.176	0.861

* = $p < 0.05$

Table O.2 Two-tailed independent sample *t*-test results (degrees of freedom (df), *t* value and *p* value) for hop height compared PRE and POST trials

Hop Height	<i>df</i>	<i>t</i>	<i>p</i>
Males	18	5.279	0.001*
Females	19	6.975	0.001*

* = $p < 0.05$

Table O.3 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), *F* value, *p* value and observed power (1- β)) to determine between sex differences over time for the hopping frequency, hopping height (ΔCOM_f) and vertical stiffness (K_{vert})

Performance	<i>df</i>	<i>F</i>	<i>p</i>	1- β
Hopping				
Frequency				
Sex * Time	3.227, 61.322	1.536	0.212	0.400
Sex	1, 19	0.040	0.843	0.054
Time	5, 95	2.486	0.037*	0.758
ΔCOM_f				
Sex * Time	5, 95	0.529	0.754	0.189
Sex	1, 19	0.980	0.335	0.156
Time	2.340, 44.455	0.828	0.460	0.194
K_{vert}				
Sex * Time	3.172, 60.273	1.143	0.341	0.301
Sex	1, 19	0.137	0.715	0.064
Time	2.795, 53.098	4.146	0.012*	0.805

* = $p < 0.05$

Table O.4 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) to determine between sex differences over time for peak vertical ground reaction force (vGRF), vertical displacement during the loading phase (ΔCOM_L) and the duration of the contact, loading, propulsion and flight phases.

Performance	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
Peak vGRF				
Sex * Time	2.887, 54.845	0.197	0.892	0.084
Sex	1, 19	0.446	0.512	0.097
Time	2.377, 45.155	1.136	0.337	0.256
ΔCOM_L				
Sex * Time	3.318, 63.038	1.098	0.360	0.297
Sex	1, 19	0.463	0.504	0.099
Time	2.935, 55.758	7.593	0.001*	0.980
Contact phase				
Sex * Time	2.599, 49.380	1.026	0.381	0.245
Sex	1, 19	1.331	0.263	0.195
Time	2.766, 52.588	9.137	0.001*	0.991
Loading phase				
Sex * Time	2.047, 38.894	0.376	0.694	0.107
Sex	1, 19	1.128	0.301	0.172
Time	2.749, 52.237	1.714	0.179	0.404
Propulsion phase				
Sex * Time	2.703, 51.349	0.474	0.682	0.135
Sex	1, 19	0.753	0.396	0.131
Time	2.535, 48.169	19.448	0.001*	1.000
Flight phase				
Sex * Time	2.942, 55.899	0.385	0.761	0.121
Sex	1, 19	1.397	0.252	0.202
Time	2.757, 52.382	8.814	0.001*	0.989

* = $p < 0.05$

Table O.5 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) to determine between sex differences over time for hip, knee and ankle joint range of motion (ROM) during loading.

Performance	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
Hip ROM				
Sex * Time	2.601, 39.158	1.653	0.204	0.332
Sex	1, 19	0.023	0.880	0.052
Time	2.467, 46.872	48.361	0.001*	1.000
Knee ROM				
Sex * Time	2.206, 41.921	0.557	0.594	0.140
Sex	1, 19	1.460	0.242	0.209
Time	2.653, 50.398	60.836	0.001*	1.000
Ankle ROM				
Sex * Time	2.370, 45.034	1.356	0.269	0.300
Sex	1, 19	0.157	0.696	0.066
Time	2.472, 46.970	1.992	0.138	0.435

* = $p < 0.05$

Table O.6 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) to determine between sex differences over time for hip, knee and ankle joint stiffness.

Performance	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
Hip stiffness				
Sex * Time	1.871, 35.549	0.568	0.560	0.134
Sex	1, 19	0.001	0.993	0.050
Time	1.990, 37.815	0.202	0.817	0.079
Knee stiffness				
Sex * Time	1.961, 37.267	0.801	0.454	0.175
Sex	1, 19	6.061	0.024*	0.647
Time	1.978, 37.578	7.262	0.002*	0.914
Ankle stiffness				
Sex * Time	1.991, 37.834	1.107	0.341	0.230
Sex	1, 19	2.508	0.130	0.324
Time	2.755, 52.349	10.832	0.001*	0.997

* = $p < 0.05$

Table O.7 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) to determine between sex differences over time for knee flexion/extension–ankle flexion/extension (KxAx), hip flexion/extension–knee flexion/extension (HxKx), knee flexion/extension–ankle abduction/adduction (KxAy) during the propulsion phase

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
KxAx				
Sex * Time	5, 95	1.008	0.417	0.346
Sex	1, 19	4.663	0.044*	0.536
Time	5, 95	8.214	0.001*	1.000
HxKx				
Sex * Time	3.230, 61.366	0.949	0.453	0.256
Sex	1, 19	1.678	0.211	0.234
Time	5, 95	26.932	0.001*	1.000
KxAy				
Sex * Time	3.063, 58.206	0.265	0.854	0.098
Sex	1, 19	6.364	0.021*	0.668
Time	3.192, 60.649	5.200	0.002*	0.922

* = $p < 0.05$

Table O.8 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) to determine between sex differences over time for knee flexion/extension–ankle flexion/extension (KxAx), hip flexion/extension–knee flexion/extension (HxKx), knee flexion/extension–ankle abduction/adduction (KxAy) couplings during the loading phase

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
KxAx				
Sex * Time	3.069, 58.309	0.313	0.820	0.108
Sex	1, 19	3.893	0.063	0.465
Time	5, 95	22.820	0.001*	1.000
HxKx				
Sex * Time	2.830, 53.774	0.692	0.553	0.183
Sex	1, 19	0.379	0.545	0.090
Time	2.822, 53.610	52.135	0.001*	1.000
KxAy				
Sex * Time	3.044, 57.828	0.371	0.777	0.119
Sex	1, 19	2.467	0.133	0.320
Time	2.788, 52.972	20.184	0.001*	1.000

* = $p < 0.05$

Table O.9 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) to determine between sex differences over time for the duration of the feedforward phase of the gluteus medius (GM), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL) and fibularis longus (FL) muscles

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM				
Sex * Time	5, 75	1.077	0.380	0.363
Sex	1, 15	0.029	0.866	0.053
Time	5, 75	0.327	0.011*	0.867
VL				
Sex * Time	5, 85	0.595	0.704	0.208
Sex	1, 17	0.706	0.412	0.706
Time	2.832, 48.150	2.720	0.058	0.608
MG				
Sex * Time	5, 75	1.342	0.256	0.449
Sex	1, 15	0.048	0.829	0.055
Time	2.320, 34.805	30.743	0.001*	1.000
SOL				
Sex * Time	5, 85	1.403	0.231	0.472
Sex	1, 17	2.016	0.174	0.268
Time	2.863, 48.674	10.117	0.001*	0.996
FL				
Sex * Time	5, 75	0.934	0.464	0.316
Sex	1, 15	0.409	0.532	0.092
Time	5, 75	5.422	0.001*	0.985

* = $p < 0.05$

Table O.10 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) to determine between sex differences over time for the total duration of gluteus medius (GM), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscle activity

Muscle	df	F	p	1-β
GM				
Sex * Time	5, 75	1.871	0.109	0.606
Sex	1, 15	0.782	0.390	0.132
Time	3.133, 46.988	1.869	0.146	0.463
VL				
Sex * Time	5, 85	2.169	0.065	0.686
Sex	1, 17	1.030	0.324	0.160
Time	5, 85	8.766	0.001*	1.000
MG				
Sex * Time	2.868, 43.020	1.025	0.389	0.254
Sex	1, 15	0.023	0.882	0.052
Time	3.016, 45.235	8.049	0.001*	0.986
SOL				
Sex * Time	3.058, 51.988	1.592	0.202	0.398
Sex	1, 17	4.951	0.040*	0.555
Time	3.110, 52.866	12.394	0.001*	1.000
FL				
Sex * Time	5, 75	0.524	0.758	0.185
Sex	1, 15	1.343	0.265	0.192
Time	5, 75	6.032	0.001*	0.993
TA				
Sex * Time	5, 30	0.928	0.477	0.284
Sex	1, 6	0.779	0.412	0.116
Time	5, 30	2.109	0.092	0.613

* = $p < 0.05$

Table O.11 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) to determine between sex differences over time for the change in MAV of the gluteus medius (GM), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL) and fibularis longus (FL) muscles during the feedforward phase

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM				
Sex * Time	5, 80	3.631	0.005*	0.909
Sex	1, 16	8.339	0.011*	0.774
Time	2.532, 40.512	6.280	0.002*	0.922
VL				
Sex * Time	1.872, 28.074	7.990	0.002*	0.922
Sex	1, 15	12.825	0.003*	0.917
Time	1.627, 24.411	23.726	0.001*	1.000
MG				
Sex * Time	2.787, 41.811	0.630	0.588	0.166
Sex	1, 15	0.730	0.406	0.126
Time	2.509, 37.630	3.432	0.033*	0.674
SOL				
Sex * Time	1.620, 27.543	1.068	0.345	0.201
Sex	1, 17	0.248	0.625	0.076
Time	1.544, 26.412	21.791	0.001*	1.000
FL				
Sex * Time	2.711, 37.955	0.439	0.707	0.159
Sex	1, 14	0.677	0.424	0.120
Time	2.266, 31.728	15.943	0.001*	1.000

* = $p < 0.05$

Table O.12 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) to determine between sex differences over time for the change in MAV of the gluteus medius (GM), biceps femoris (BF), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscles during the background phase

Muscle	df	F	p	1-β
GM				
Sex * Time	5, 90	1.670	0.150	0.556
Sex	1, 18	2.901	0.106	0.364
Time	5, 90	6.353	0.001*	0.995
BF				
Sex * Time	2.906, 52.309	1.479	0.232	0.363
Sex	1, 18	0.000	0.996	0.050
Time	3.056, 55.008	7.405	0.001*	0.980
VL				
Sex * Time	1.980, 35.636	6.505	0.004*	0.879
Sex	1, 18	18.155	0.001*	0.980
Time	2.366, 42.588	15.482	0.001*	1.000
MG				
Sex * Time	2.579, 43.851	1.046	0.374	0.247
Sex	1, 17	0.050	0.826	0.055
Time	2.673, 45.439	1.985	0.136	0.450
SOL				
Sex * Time	1.733, 31.188	3.688	0.042*	0.595
Sex	1, 18	5.141	0.036*	0.574
Time	1.636, 29.447	24.845	0.001*	1.000
FL				
Sex * Time	2.776, 47.195	0.350	0.774	0.111
Sex	1, 17	0.863	0.366	0.142
Time	2.383, 40.518	6.768	0.002*	0.931
TA				
Sex * Time	2.386, 40.560	0.264	0.806	0.092
Sex	1, 17	0.930	0.348	0.149
Time	1.782, 30.294	1.953	0.163	0.353

* = $p < 0.05$

Table O.13 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) to determine between sex differences over time for the change in MAV of the gluteus medius (GM), biceps femoris (BF), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscles during the M1 phase

Muscle	df	F	p	1-β
GM				
Sex * Time	3.133, 56.393	0.818	0.494	0.220
Sex	1, 18	0.635	0.436	0.117
Time	5, 90	9.377	0.001*	1.000
BF				
Sex * Time	5, 90	0.302	0.911	0.122
Sex	1, 18	0.976	0.336	0.155
Time	2.334, 42.008	11.761	0.001*	0.996
VL				
Sex * Time	1.602, 28.828	1.439	0.252	0.257
Sex	1, 18	4.307	0.053	0.502
Time	1.599, 28.782	24.153	0.001*	1.000
MG				
Sex * Time	2.334, 39.680	0.533	0.618	0.138
Sex	1, 17	0.536	0.474	0.106
Time	5, 85	8.604	0.001*	1.00
SOL				
Sex * Time	2.153, 38.757	0.455	0.455	0.185
Sex	1, 18	0.142	0.238	0.212
Time	2.739, 49.299	61.295	0.001*	1.000
FL				
Sex * Time	2.705, 45.982	0.907	0.437	0.223
Sex	1, 17	3.358	0.084	0.409
Time	2.110, 35.867	11.774	0.001*	0.993
TA				
Sex * Time	1.958, 33.288	0.139	0.867	0.069
Sex	1, 17	0.261	0.616	0.077
Time	1.729, 29.401	3.589	0.046*	0.580

* = $p < 0.05$

Table O.14 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) to determine between sex differences over time for the change in MAV of the gluteus medius (GM), biceps femoris (BF), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscles during the M2 phase

Muscle	df	F	p	1-β
GM				
Sex * Time	2.790, 50.221	0.539	0.645	0.149
Sex	1, 18	0.177	0.679	0.068
Time	5, 90	32.543	0.001*	1.000
BF				
Sex * Time	2.606, 46.905	2.581	0.072	0.559
Sex	1, 18	3.653	0.072	0.440
Time	5, 90	16.401	0.001*	1.000
VL				
Sex * Time	2.307, 41.527	12.362	0.001*	0.997
Sex	1, 18	23.896	0.001*	0.996
Time	2.558, 46.041	17.889	0.001*	1.000
MG				
Sex * Time	5, 85	1.816	0.118	0.595
Sex	1, 17	0.735	0.403	0.128
Time	2.823, 47.997	5.433	0.003*	0.905
SOL				
Sex * Time	2.610, 46.987	1.444	0.244	0.334
Sex	1, 18	2.052	0.169	0.274
Time	2.578, 46.410	25.075	0.001*	1.000
FL				
Sex * Time	5, 85	2.355	0.047*	0.728
Sex	1, 17	2.481	0.134	0.318
Time	2.018, 34.306	8.829	0.001*	0.959
TA				
Sex * Time	2.742, 46.614	1.025	0.385	0.250
Sex	1, 17	0.195	0.664	0.070
Time	2.145, 36.463	5.401	0.008*	0.931

* = $p < 0.05$

Table O.15 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) to determine between sex differences over time for the change in MAV of the gluteus medius (GM), biceps femoris (BF), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscles during the M3 phase

Muscle	df	F	p	1-β
GM				
Sex * Time	5, 90	1.023	0.409	0.350
Sex	1, 18	5.430	0.032*	0.597
Time	2.594, 46.695	18.368	0.001*	1.000
BF				
Sex * Time	3.066, 55.182	4.468	0.007*	0.861
Sex	1, 18	11.849	0.003*	0.902
Time	2.565, 46.167	14.335	0.001*	1.000
VL				
Sex * Time	2.532, 45.583	5.555	0.004*	0.888
Sex	1, 18	13.188	0.002*	0.929
Time	1.731, 31.161	21.016	0.001*	1.000
MG				
Sex * Time	5, 85	2.774	0.023*	0.806
Sex	1, 17	3.754	0.069	0.448
Time	3.009, 52.684	2.626	0.058	0.620
SOL				
Sex * Time	2.587, 46.574	2.662	0.067	0.571
Sex	1, 18	7.393	0.014*	0.730
Time	2.392, 43.048	13.701	0.001*	0.999
FL				
Sex * Time	5, 85	1.166	0.333	0.396
Sex	1, 17	4.476	0.049*	0.514
Time	2.771, 47.103	4.767	0.007*	0.856
TA				
Sex * Time	2.539, 43.155	1.821	0.165	0.405
Sex	1, 17	3.553	0.077	0.428
Time	2.206, 37.503	8.204	0.001*	0.959

* = $p < 0.05$

Table O.16 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1-β)) to determine between sex differences over time for the change in MAV of the gluteus medius (GM), biceps femoris (BF), vastus lateralis (VL), medial gastrocnemius (MG), soleus (SOL), fibularis longus (FL) and tibialis anterior (TA) muscles during the propulsion phase

Muscle	<i>df</i>	<i>F</i>	<i>p</i>	<i>1-β</i>
GM				
Sex * Time	2.007, 36.117	1.703	0.196	0.335
Sex	1, 18	4.161	0.056	0.488
Time	2.754, 49.574	29.775	0.001*	1.000
BF				
Sex * Time	2.66, 47.997	3.434	0.028*	0.702
Sex	1, 18	11.460	0.003*	0.893
Time	1.734, 31.217	6.484	0.006*	0.843
VL				
Sex * Time	1.709, 30.765	1.017	0.363	0.199
Sex	1, 18	0.872	0.363	0.143
Time	1.998, 35.959	28.415	0.001*	1.000
MG				
Sex * Time	2.352, 39.982	1.849	0.165	0.393
Sex	1, 17	3.355	0.085	0.408
Time	2.306, 39.205	9.318	0.001*	0.981
SOL				
Sex * Time	2.583, 46.491	0.531	0.637	0.143
Sex	1, 18	1.501	0.236	0.213
Time	2.002, 36.043	7.390	0.002*	0.920
FL				
Sex * Time	2.264, 38.493	1.379	0.265	0.295
Sex	1, 17	3.502	0.079	0.423
Time	5, 85	2.534	0.035*	0.764
TA				
Sex * Time	2.161, 36.735	0.332	0.736	0.101
Sex	1, 17	0.223	0.643	0.073
Time	2.579, 43.842	19.298	0.001*	1.000

* = $p < 0.05$

Table O.17 6 (time period) x 2 (sex) repeated measure ANOVA results (degrees of freedom (df), f value, p value and observed power (1- β)) to determine between sex differences over time for the co-activation of the vastus lateralis and biceps femoris (VL:BF) muscle pair and the soleus and tibialis anterior (SOL:TA) muscle pair during each time phase

Muscle	df	F	p	1-β
Background				
VL:BF				
Sex * Time	2.748, 43.975	2.682	0.067	0.579
Sex	1, 16	1.834	0.194	0.247
Time	2.414, 38.619	0.338	0.755	0.104
SOL:TA				
Sex * Time	1.885, 28.275	2.425	0.109	0.435
Sex	1, 15	4.401	0.053	0.501
Time	2.711, 40.661	12.983	0.001*	0.999
M1				
VL:BF				
Sex * Time	2.821, 45.410	1.411	0.253	0.339
Sex	1, 16	3.173	0.094	0.388
Time	5, 80	1.752	0.132	0.576
SOL:TA				
Sex * Time	2.168, 32.519	2.380	0.105	0.464
Sex	1, 15	1.903	0.188	0.253
Time	2.758, 41.368	8.811	0.001*	0.987
M2				
VL:BF				
Sex * Time	1.778, 28.450	1.233	0.303	0.235
Sex	1, 16	2.301	0.149	0.297
Time	2.010, 32.161	1.369	0.269	0.274
SOL:TA				
Sex * Time	5, 75	2.570	0.034*	0.766
Sex	1, 15	1.881	0.190	0.250
Time	2.508, 37.621	2.787	0.063	0.575
M3				
VL:BF				
Sex * Time	2.224, 35.583	0.758	0.489	0.175
Sex	1, 16	3.125	0.096	0.383
Time	5, 80	1.962	0.093	0.633
SOL:TA				
Sex * Time	2.549, 38.231	2.089	0.126	0.454
Sex	1, 15	1.497	0.240	0.209
Time	1.931, 28.962	1.211	0.311	0.407

* = $p < 0.05$