

School of Physiotherapy and Exercise Science

**An investigation into perception of change in the foot-floor interface
during repeated stretch-shortening cycles**

Mervyn Joseph Travers

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Doctor of Philosophy

of

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Declaration

To the best of my knowledge and belief this thesis contains no material previously published by any other person except where due acknowledgement has been made.

This thesis contains no material which has been accepted for the award of any other degree or diploma in any university.

Mervyn Joseph Travers

Signature:

Date:

Abstract

Proprioception is widely considered to be a critical sensory input for safe and efficient motor planning and execution of movement. Most of the current knowledge base on proprioception has been obtained from testing position matching, movement detection and force matching at isolated joints. These precise open-chain tests have allowed us to understand the subtleties of the peripheral inputs which contribute to one's conscious sensations of position or movement. However, it is difficult to draw inference from such methodologies to more dynamic function such as locomotion which is governed by repeated stretch shortening cycles (SSC) of the lower limb. To date, there is an absence of any testing paradigm that measures participants' proprioceptive function during a SSC task.

In biomechanical studies, locomotion is commonly modelled using a spring mass representation which describes the lower-limb behaving like a spring to progress the centre of mass. During locomotion the foot/floor interface is rarely homogenous, and we cater for such external challenges by modulating the stiffness of the 'leg-spring' using a combination of feedforward and feedback control. These minor external perturbations are usually accommodated without conscious perception, yet a threshold exists where we are alerted to the challenge / resulting change in configuration. The current proprioception model is based on the acute sense of one's own self, but it may be that one's ability to detect changes in our external environment might also be relevant.

As the ankle complex is considered a primary driver of low load SSC function in the lower limb it is the focus of this body of work. The first step in this thesis was to investigate the reliability of a custom built sleigh system as an appropriate environment for investigating the repeated SSC of the ankle. This was achieved by comparing the hopping performances of 13 healthy participants, as measured by our kinematic derived variables under upright and sleigh conditions. The results suggested that sleigh hopping represents a highly reliable and appropriate task for examination of ankle stiffness and kinematic variables on a within (all ICC > .975) and between day (all ICC > .852) basis whilst negating the confounding features of upright hopping such as fatigue and balance.

This thesis aimed to merge the experimental paradigms of proprioceptive testing and the examination of lower limb stiffness modulation. To this end, the novel Minimal Perceptible

Difference (MPD) protocol has been developed and represents a shift in research focus towards investigating proprioception under more functional circumstances by utilising a sleigh hopping model to measure (in mm) participants' ability to detect changes in the floor surface height during repeated SSC. The next phase in this thesis was therefore to introduce and establish the theoretical basis for the MPD test and to examine the reliability of the MPD test in a normal population ($n = 16$) on a within and between day basis. The results demonstrated the MPD test is a reliable tool for assessing participants' ability to detect changes in the floor surface height during repeated SSC. Consistent with observations from existing proprioception research, statistically significantly greater sensitivity was observed when performing the MPD test using a bilateral hopping technique (MPD = 15.7mm; 95% CI 14.1–17.2 mm) when compared to alternating foot strikes (MPD = 26.6mm; 95% CI 25.1–28.1mm; $p < .001$). This may reflect finely tuned sensory requirements for upright stance which may be much less relevant for (bipedal) gait further questioning the utility of traditional proprioception testing in dynamic tasks.

Based upon the methodology established by the MPD, the next phase of the thesis was to investigate the effect of expectation of changes in the hopping environment on participants' subsequent motor strategies. It was demonstrated that expectation of a change in the foot / landing surface interface produced a change in baseline bilateral hopping strategy. These changes included reduced contact time, increased stretch velocity bilaterally and an increase in stiffness in the leg on the side of the expected change and EMG signal amplitude increased in both agonist and antagonist muscles of the ankles (all $p < .05$). No other studies have demonstrated such feedforward responses in a risk minimised environment as per the MPD protocol.

Accounting for the role of expectation during the testing protocol, this thesis proceeded to explore the differences in motor strategy response to perceived and unperceived perturbations during hopping. Furthermore, this body of work compared the differences between the expectation of a change in the environment, the perception of the change and the experience of a subliminal perturbation on participants' motor behaviour

Traditional proprioceptive testing methods specifically utilise measures of range or position to quantify proprioceptive function. However, the combined findings of these studies

demonstrate minimal differences in range or position related outputs for the ankle when either subliminal or perceptible perturbations were introduced (all $p > .05$). Under the present experimental circumstances, expectation of a change in the foot-floor interface generated significant changes in motor behaviour, whereas the introduction of either perceived or subliminal floor height changes did not. These findings further question the utility of traditional testing methods of measuring proprioception in the context of dynamic function.

Proprioceptive input is critical for normal and safe movement. However, there exists a gap in the literature regarding the assessment of proprioceptive function during dynamic tasks of the lower limb. This doctoral research aimed to address this issue and serves as a foundation for considering proprioception as it pertains to dynamic function at the ankle. Further research is required in this area and the avenues for further investigation are considered and discussed in this thesis.

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‘Not all those who wander are lost’ – J.R.R. Tolkien

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Statement of Contribution

The author contributed to the development of the research idea, methods, pilot testing, recruitment, data collection, data management, data analysis and preparation of all chapters within this manuscript and the above publications.

List of Abbreviations

BH	Baseline Hopping Strategy
COM	Centre of Mass
EH	Hopping with expectation of a perturbation
K	Leg Stiffness
M	Body Mass
MPD	Minimal Perceptible Difference
P	Perceived perturbation during hopping
ROM	Range of Motion
SH	Sleigh hopping
SSC	Stretch-shortening cycle
SubP	Subperceptual perturbation during hopping
t_c	Contact Time
t_f	Flight Time
UH	Upright hopping
θ_{contact}	Ankle angle at contact in sagittal plane
θ_{peak}	Peak dorsiflexion angle
$\theta_{\text{c-p}}$	Stretch Amplitude: the change in the ankle joint angle from landing (contact) to the most dorsiflexed point
θ_{vel}	Stretch Velocity (θ_{vel})
$\theta_{\text{take-off}}$	Ankle angle at take-off θ in sagittal plane

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1. Review of the literature and rationale for thesis

1.1 Introduction

The concept of our conscious perception of movement has long been a focus of scientific interest and in 1826 Charles Bell was the first to attribute an anatomical basis for movement sense, body position and awareness of contraction. He proposed that between the brain and the muscles there exists a “circle of nerves” with one nerve conveying messages from the brain to the muscle and the other nerve informing the brain of the condition of that muscle (Bell 1826).

In 1906, Sherrington described proprioception as afferent information from proprioceptors located in the proprioceptive fields that contribute to conscious sensations, total posture and segmental posture (Sherrington 1906). This definition of proprioceptors related specifically to all subcutaneous, musculoskeletal structures giving rise to sensations of segmental position, movement, force, weight, effort, pressure, body position / posture and balance. He also categorised and described other somatic senses arising from the exteroceptors of the skin allowing for the transmission of information pertaining to touch, pain, pressure, vibration, temperature and deep tissue receptors also carrying signals of temperature and pain (Sherrington 1906). Thus, the somatic sensory system is suggested to give us a “body sense” and proprioceptors are the specialised receptors which specifically inform us about how our body is positioned and moving in space (Bear, Connors et al. 2007). As research in this field has evolved over the last century it appears that terms such as proprioception, joint position sense and kinaesthesia have become synonymous and thus have been used interchangeably (Stillman 2002). Therefore in the interest of clarity, this thesis will discuss proprioception in terms of the Sherrington’s broad definition which includes its constituent parts such a movement sense, sense of effort, postural equilibrium and position sense.

The concept of proprioception is central to this thesis; however the focus is not the basic physiological processes involved. Although the current chapter is not intended to be an all-inclusive review of the basic physiological processes governing proprioception, the pertinent literature which has evolved the current model of proprioception will be discussed. An excellent review of proprioceptive research in its entirety has been recently published (Proske and Gandevia 2012). The intention of this chapter is to outline the gap between information gleaned from the mechanistic science and the likely normal operation of the lower limb. Furthermore, it is intended to establish an argument for a shift in research focus

towards investigating proprioception under more functional circumstances. Although this thesis is focused on the area of the ankle, the majority of proprioceptive research has been conducted elsewhere. For example, the upper limb has been subject to particular attention. However, literature relevant to the foot and ankle complex and in particular movements in the sagittal plane (plantarflexion / dorsiflexion) will be expanded upon.

1.2 Vibration studies

The early part of the 20th century was dominated by discussion as to which peripheral receptors were responsible for proprioception and there was division over the roles of joint receptors and muscle spindles in particular. A major breakthrough occurred in a series of experiments which demonstrated the effects of muscle vibration on participants' sense of position and sense of movement (Goodwin, McCloskey et al. 1972; Goodwin, McCloskey et al. 1972). Vibration disrupts proprioceptive signalling via stimulation of type Ia sensory fibres – a nerve fibre which signals length changes of the muscle spindle (Bear, Connors et al. 2007). They observed movement tracking errors of up to forty degrees at the elbow during vibration to the biceps or triceps muscles (Goodwin, McCloskey et al. 1972). The role of intramuscular receptors was further emphasised when participants could detect both passive and active movement and repositioning of the finger post anaesthetic injection intended to negate influence of joint receptors (Goodwin, McCloskey et al. 1972). Though this did not definitively refute the role of other peripheral receptors, the overwhelming evidence of intramuscular contribution then focused future research to examine this phenomenon further. For example, diminished passive movement discrimination of the distal interphalangeal joint when the tendons had been disengaged by hand positioning further demonstrated the relevance of intramuscular signals (Gandevia and McCloskey 1976).

Any doubt remaining regarding intramuscular signals and their role in proprioception were allayed in an experiment where participants during surgery under local anaesthetic had stretches applied to exposed wrist and hand tendons. They detected the stretches imposed on their muscles, and reported them as rotations of their relevant joints (McCloskey, Cross et al. 1983). The same authors demonstrated in a single-participant experiment comparable acuity for the detection of stretch imposed on an exposed flexor hallucis longus tendon as for the intact toe. Furthermore, this participant experienced illusory movements of plantarflexion of the big toe when vibration was applied directly to the exposed flexor hallucis longus tendon (McCloskey, Cross et al. 1983). These vibration-induced illusions were consistent with those

reported by Goodwin et al. (1972) for vibration applied through the skin to the biceps and triceps brachii tendons. Finally, it appears that the magnitude of the vibration-induced illusory movement is proportional to the amplitude and frequency of the vibration stimulus (Goodwin, McCloskey et al. 1972; McCloskey 1973) and the performance errors are largest when multiple synergists are stimulated (Verschueren, Cordo et al. 1998). Thus by using vibration as to simulate 1a afferents, intramuscular receptors have been strongly implicated in the existing literature as key components in the detection of movement and position.

1.3 Task dependent utility of proprioceptive information

In the above instances vibration stimulus was utilised to elicit illusory movements during isolated single joint movements such as isolated elbow flexion, finger flexion or toe flexion. However, humans rarely perform these movements in isolation so more recently the literature has raised more applied questions such as; what happens if vibration is applied to participants during normal locomotion? Ivanenko et al (2000) examined the effect of leg muscle vibration on human walking. They applied vibration to the quadriceps, hamstring, triceps surae and tibialis anterior muscles of both legs and observed the effect on the key functional tasks of the lower limb - quiet stance and walking. In stance, they observed that tibialis anterior vibration resulted in a forward body tilt (Ivanenko, Grasso et al. 2000). Hamstring vibration resulted in backwards trunk lean, whereas triceps surae vibration induced a whole body inclination backwards (Ivanenko, Grasso et al. 2000). It is evident that the illusory movement effect resulted in motor responses which were specific to the location on the leg of the applied stimulus. Such vibration induced postural tilting represents a compensatory response to illusory lengthening of the vibrated muscle and has also been observed via application of unilateral vibration of both leg, abdominal and neck muscles during stance (Courtine, De Nunzio et al. 2007).

1.4 Passive testing methodologies

Importantly during locomotion, vibration results in a very different response than upright stance. For example during forward walking, hamstring vibration produced significantly larger increases in walking speed compared to vibration of any other leg muscles (Ivanenko, Grasso et al. 2000). Conversely, the same hamstring stimulus produced a slowing of reverse walking, but quadriceps vibration increased the speed of backwards locomotion (Ivanenko, Grasso et al. 2000). Interestingly, application of unilateral vibration to the abdominal, trunk or neck muscles has been shown to induce significant changes in the trajectory of gait away from the side of the stimulus whereas unilateral leg muscle vibration causes little to no

change in trajectory of gait (Courtine, De Nunzio et al. 2007). These findings demonstrate that in the face of the same stimulus the gating or utility of proprioceptive information is highly task dependent (Ivanenko, Grasso et al. 2000). In particular, it appears the motor responses to the same stimulus are very different for quiet stance than for locomotion. This raises the question; if the way we use proprioceptive information is highly task dependent, then should proprioceptive tests be aligned with functional tasks?

Although passive movements are not a common occurrence in humans' day to day activities, a large proportion of the existing literature on the sensorimotor system has been obtained through the study of passive movement detection and position matching tasks. Movement detection methodologies involve open-chain movements being externally applied to a passive limb via a servomotor and participants are instructed to indicate the perception of movement, the direction of movement or both (See Figure 1.1 for an illustration of the experimental set-up of Wise et al (1998) which is representative of the methodologies employed in the wider literature). Although a slightly different apparatus may be required based on the configuration of the different joints in the body, the same experimental model has been applied to the upper limb (Gandevia and McCloskey 1976; Wise, Gregory et al. 1996; Wise, Gregory et al. 1998) and lower limb, including the ankle (Refshauge, Chan et al. 1995; Refshauge and Fitzpatrick 1995; Matre, Arendt-Neilsen et al. 2002). Position matching tasks are similarly common in the literature and are considered a measure of position sense (Feuerbach, Grabiner et al. 1994; Matre, Arendt-Neilsen et al. 2002; Vuillerme, Chenu et al. 2006; Allen, Ansems et al. 2007; Vuillerme and Cuisinier 2008; Allen, Leung et al. 2010). Participants are usually seated with and blind-folded for such tests. The joint of interest on one of their limbs is placed in a particular position by the testers e.g. 30 degrees of elbow extension (typically referred to as the reference limb). They are then required to match this position with the corresponding joint on the other limb (typically termed the indicator limb). See Figure 1.2 for an illustration of a position matching testing protocol at the elbow and knee which is representative of those widely applied in research settings.

As stated previously, it is clear that muscles play a significant role in the detection of passive movement and position. In particular, it appears a muscle's resting length prior to passive movement testing can largely influence participants detection thresholds (Wise, Gregory et al. 1996). This is likely due to the passive mechanical effect on the muscle fibres (Proske, Morgan et al. 1993) and more specifically the intrafusal fibres of the muscle spindle (Proske,

Wise et al. 2000). Furthermore, the contraction history of the muscle is equally relevant to participants' proprioceptive acuity. Consistent inter-participant variability was commonly reported in the literature examining detection of passive movement and a suggested cause was variations in fusimotor tone (Gandevia and McCloskey 1976). It was later demonstrated that conditioning contractions of the elbow flexors allowed for significantly more sensitive detection of passive movement into extension at the elbow (Wise, Gregory et al. 1996). A similar response was observed when the extensors were exposed to a conditioning contraction prior to passive movement detection testing (Wise, Gregory et al. 1996). As such, subsequent methodologies incorporated conditioning contractions to negate slackening of the muscle spindle as a confounding factor (Wise, Gregory et al. 1998) (See figure 1.1 A&B).

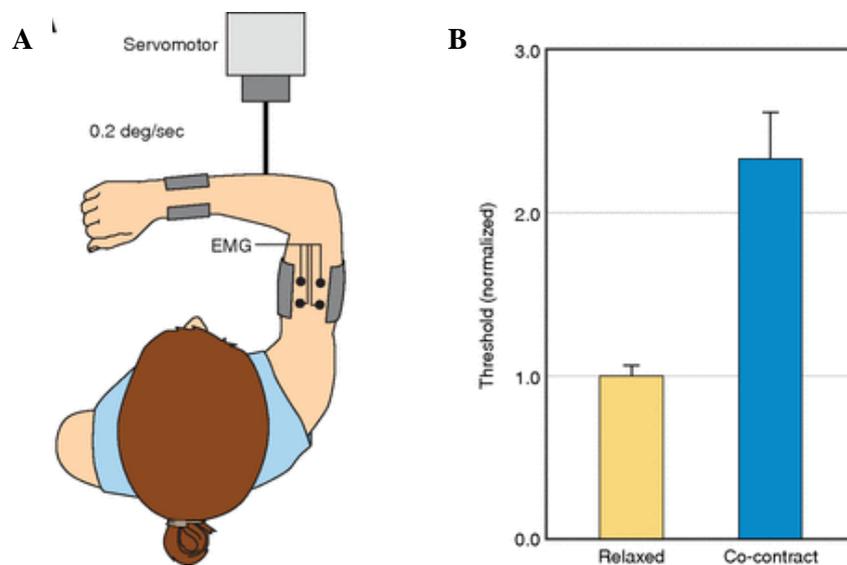


Figure 1.1 Measurement of passive movement detection thresholds

A: blindfolded subjects were required to indicate the direction of small movements (0.2°/s) applied to the right forearm with a servomotor. Detection thresholds were measured for elbow extension and flexion movements under relaxed and co-contraction conditions (15% MVC co-contraction of elbow flexors and extensors, monitored as EMG). B: average thresholds measured for 7 subjects. There were no differences between thresholds for flexion and extension movements, and values have been pooled. Thresholds have been normalized with respect to the average threshold measured for the relaxed condition. Thresholds measured during co-contractions of elbow muscles (blue bar) were significantly higher than when the arm was relaxed (orange bar). Images and caption reproduced from Proske and Gandevia (2012).

The functional meaning of the above is that the muscle spindle is much less likely to become slack at longer muscle lengths. This was demonstrated at the ankle where movement detection thresholds were three times lower in standing than in sitting where a bent knee put

the gastrocnemius muscle at a shorter operating length (Refshauge and Fitzpatrick 1995). All information regarding the proprioceptive system to this point had been obtained by studying the upper limb. Therefore, that study was the seminal investigation of the proprioceptive function of the lower limb and also the consideration of the role of muscle activity in the detection of passive movements by placing the participants in standing / weight bearing. When the participants were seated with their ankles replicating the standing position they were equally sensitive to passive movement as when standing (Refshauge and Fitzpatrick 1995). Refshauge et al (1995) observed a mean passive movement detection threshold into ankle dorsiflexion of 0.66 (+/-0.09) degrees at 0.05 degrees/s with their participants seated with their knees bent. This reduced to 0.24 (+/-0.03) degrees with the knees straightened and reduced again to 0.20 (+/-0.04) in standing. Movement detection thresholds of below 1 degree have also been demonstrated using a similar test at speeds of 5 degrees/s and 10 degrees/s into both ankle dorsiflexion and plantarflexion (Matre, Arendt-Neilsen et al. 2002). It is difficult to compare these thresholds directly as it has been shown that such thresholds are highly velocity dependent and can be reduced up to tenfold with increased passive movement velocity (Hall and McCloskey 1983; Refshauge, Chan et al. 1995). Conversely, if the stretch rate is sufficiently slow a velocity independent relationship becomes evident and this is considered to be a reflection of position sense rather than movement detection (Clark, Burgess et al. 1985). Nonetheless, such remarkable acuity suggests that individuals are extremely sensitive at consciously perceiving passive movement in the sagittal plane at the ankle joint. Consistently, Blaszczyk et al (1993) observed matching errors of less than 1 degree in both ankle plantarflexion and dorsiflexion and suggested that the high degree of accuracy in their test maybe due to the perception of gravitationally induced forces resulting from the upright position (Blaszczyk, Hansen et al. 1993). The literature suggests that this is a reasonable conclusion and is substantiated in part by other research which has demonstrated that muscle spindle signals and sense of effort combine to optimise sense of limb position during active movement (Winter, Allen et al. 2005).

An underlying assumption in the wider literature is that it is optimal to be more sensitive during proprioceptive testing. This may be true when considering upright stance and maintenance of posture where it is clear that healthy people exhibit extremely high sensitivity to sagittal plane ankle position during upright stance (Blaszczyk, Hansen et al. 1993). It is consistent therefore, that acuity declines with aging along with other features considered integral to maintenance of upright posture such as maximum voluntary excursion of the centre of mass and postural sway (Blaszczyk, Hansen et al. 1993; Blaszczyk, Lowe et al. 1993; Blaszczyk, Lowe et al. 1994; McClenaghan, Williams et al. 1995). However, given

the 'strategy dependent gating' of proprioceptive information demonstrated by comparing the effect of vibrations on both stance and locomotion (Ivanenko, Grasso et al. 2000; Courtine, De Nunzio et al. 2007) it is plausible that this high level of acuity may not be equivalent between the two tasks. Given the absence of an appropriate testing methodology to measure proprioception during locomotion, this hypothesis has not been explored to date.

The finely tuned perception of sagittal passive movement at the ankle has been demonstrated to be remarkably robust in the presence of pain. Only one paper has explored this specifically. Matre et al (2002) compared the effect of hypertonic saline injection (6%) into the soleus and tibialis anterior muscles individually and to both muscles simultaneously on sagittal plane passive movement detection at the ankle. They compared the same protocol using an isotonic saline injection (0.9%). The isotonic saline injections induced no pain in the participants and their movement detection was unaffected (Matre, Arendt-Neilsen et al. 2002). Furthermore, no effect was observed when the pain-inducing hypertonic saline injections were administered to the muscles individually. However, hypertonic saline injections administered to both muscles simultaneously did reduce their participants' ability to detect passive movement (Matre, Arendt-Neilsen et al. 2002). The fact that it required an experimental pain injection into both agonist and antagonist muscles demonstrates that the proprioceptive system is resilient to muscle pain when it comes to movement detection. Conversely, the effect of local muscle fatigue on movement detection is more significant. For example, it has been demonstrated that local muscle fatigue reduced participants' ability to distinguish passive movement of the shoulder at varying speeds (Pedersen, Lonn et al. 1999). However, the impact of fatigue seems much more relevant to tests of joint position and active movement (Walsh, Hesse et al. 2004; Allen, Leung et al. 2010; Fortier and Basset 2012) and sense of effort (Proske and Gandevia 2012).

Although the movement detection and position matching tests used in such studies have produced useful information and contributed to the wider understanding of proprioception, it has been acknowledged by prominent researchers in this field that they do not necessarily replicate normal function (Proske, Wise et al. 2000). Furthermore, whilst findings relevant to contraction history, conditioning, muscle spindle tone have contributed to understanding of the physiology of proprioception it is plausible that such fine properties may not be relevant to dynamic tasks such as performance of repeated SSC. A similar hypothesis has been explored previously where the subtleties of the stretch reflex (SR) of the soleus muscle observed during controlled experimental testing may become less significant during more

functional tasks. Traditionally, the SR is considered an important component in the modulation of ankle stiffness during the normal function of the lower limb – walking (Sinkjær, Andersen et al. 1996), landing (Voigt, Dyhre-Poulsen et al. 1998) and hopping (Funase, Higashi et al. 2001). Research in this area has been robust and precise, utilising participants with complete spinal cord injury (unlikely supraspinal influence) to demonstrate the modulation of the SR via alteration of hip position (peripheral influence) (Kawashima, Yano et al. 2006). However, a recent paper demonstrated that altering hip position had no observable modulating effect on the SR during repeated hopping (Gibson, Campbell et al. 2013). This demonstrates how the subtleties of a physiological process yielded from controlled experimental designs may become less significant when applied to a more functional setting. It is similarly plausible that during repeated SSC those subtleties such as contraction history, slack length and the finely tuned acuity of less than 1 degree for movement detection threshold into ankle dorsiflexion may be less relevant than during single repetition position matching tasks. Once again, this hypothesis has not been explored with respect to proprioception during locomotion due to the absence of an appropriate methodology for examining proprioception during repeated SSC.

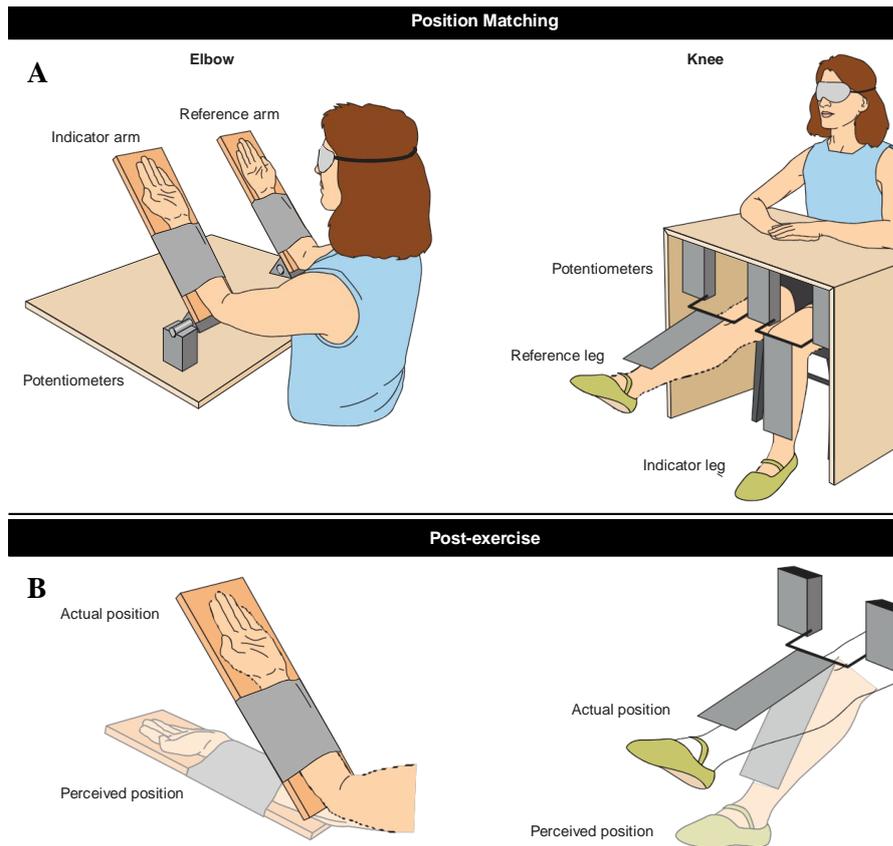


Figure 1.2 Examples of position matching tasks at the knee and elbow

A illustrates a common testing methodology for position matching of the elbow and knee. B demonstrates the directional effect of fatigue on perceived position - the exercised elbow was perceived as being more extended than it was whilst conversely, the exercised knee was perceived as being more flexed than it was. Image reproduced from (Proske and Gandevia 2012)

1.5 The influence of muscle activity on proprioception

A fundamental barrier to directly answering whether contraction enhances proprioceptive feedback is that during active movement the input to muscle spindles changes which in turn influences their respective outputs (Windhorst 2007). It has been demonstrated that active contraction of the finger flexors improved the detection of movement at the distal interphalangeal joint of the finger (Gandevia and McCloskey 1976). The argument in favour of increased acuity during active contraction was further supported when movement detection thresholds were decreased during active flexion of the elbow (Colebatch and McCloskey 1987). Countering these findings, Wise et al (1998) observed more acute detection of motion at the elbow during passive testing compared to during active contraction. An important consideration was that they specifically controlled for the effect of spindle slackening by including conditioning contraction in their methodology prior to the

passive testing (Wise, Gregory et al. 1998). Another consideration is that Wise et al (1998) instructed their participants to co-activate the biceps and triceps during testing (See figure 1.1). It has been suggested that movement sense at the ankle may be determined by the difference in firing rates between the shortening agonist and the stretched antagonist generating information regarding movement velocity (Ribot-Ciscar and Roll 1998). Applying this consideration in this case it is plausible that the co-activation of the biceps and triceps may have negatively influenced the threshold of detection for movement of the elbow (Proske, Wise et al. 2000). It may also be that in the case of co-activation there may be “more noise in the system” making detection of movement less sensitive (Proske, Wise et al. 2000).

Adding to the complexity, muscle contraction will invariably elicit some Golgi tendon organ (GTO) discharge as they respond to the force exerted upon them by connecting muscle fibres (Windhorst 2007). Although they respond to stretch, GTO are considered to primarily operate as contraction receptors (Proske and Gandevia 2012) signalling small changes in contractile force (Jami 1992). Their architecture is such that every motor unit in a single muscle is connected to at least one GTO so that information can be provided from subunits of a muscle or from the muscle in its entirety via combinations of GTO (Proske and Gandevia 2012). However, their contribution cannot be experimentally separated from other sources of proprioception so it is difficult to quantify their contribution in isolation during movement (Jami 1992). Furthermore, the responses of GTO to low level contractions are highly dependent on the preceding contraction history (Thompson, Gregory et al. 1990). Specifically, Thompson et al (1990) demonstrated large force matching errors after the completion of a maximal voluntary contraction of the elbow flexors. Therefore, as previously demonstrated with muscle spindles, the literature demonstrates that contraction history has a significant impact on the sense of tension and thus on proprioceptive signalling.

The weight of the limb itself and the sense of effort are essential components of the combined proprioceptive information associated with movement. Using position matching at the elbow, it has been demonstrated that supporting / unweighting the ‘reference arm’ rendered participants less accurate at matching its position with the other arm (Winter, Allen et al. 2005). Furthermore, the addition of a 2kg weight to the reference arm produced matching errors which were consistent with previous errors demonstrated in the presence of fatigue and thought to be linked to sense of effort (Walsh, Hesse et al. 2004; Winter, Allen et al. 2005). Whether sense of effort contributes to movement sense was ruled out when

fatiguing elbow flexors resulted in position matching errors, but did not affect movement tracking adversely. Thus, sense of effort as its own independent entity is considered a significant factor in active tests of proprioception which likely represents a function of the CNS (Winter, Allen et al. 2005; Proske and Gandevia 2012).

With repeated active movement, fatigue is an important consideration and its effect on proprioception has been widely investigated. The particular effects of fatigue and resultant weakness has been suggested as a risk factor for injury in athletes (Orchard 2002) and also as a risk factor for falls in elderly populations (Butler, Lord et al. 2008). It has been demonstrated that exercise of a particular muscle group, leading to a 30% fall in force resulted in significant position-matching errors at the relevant joint (Allen, Ansems et al. 2007) which was consistent with the findings of other studies (Saxton, Clarkson et al. 1995; Lattanzio, Petrella et al. 1997; Fortier, Basset et al. 2010). The literature suggests that feedback from the fatigued muscle is interpreted as the muscle being longer i.e. fatiguing the elbow flexors resulted in the sensation of the arm being more extended (See Figure 1.2). This was demonstrated at both the elbow and at the knee where quadriceps fatigue resulted in a sense of the knee being more flexed (Allen, Ansems et al. 2007). However, a follow up series of experiments revealed that the antagonist muscles acting at a joint produces position-matching errors in the same directions i.e. fatiguing the triceps also resulted in participants perceiving their elbow to be more extended (Allen, Leung et al. 2010). Likewise, the participants perceived their knee to be more flexed post hamstring fatigue (Allen, Leung et al. 2010). Whilst it is acknowledged that fatigue may affect the peripheral receptors such as muscle spindles, it has been suggested that it is likely that the findings above are driven by changes in the CNS (Allen, Leung et al. 2010).

In summary, there are specific peripheral receptors which contribute to our sense of position and movement during active movements. Much like in the case of passive movement, acuity in the relevant tests is affected by peripheral subtleties such as the contraction history of the muscle. It also seems that central factors such as the sense of effort contribute to proprioceptive feedback in active movement. It is evident that all such signals are highly susceptible to fatigue effects. Therefore when devising a proprioceptive test based on dynamic function it would be important to ensure that fatigue was negated as a confounding factor. This shall be explored further in Chapter 3 – Upright vs Sleight Hopping.

1.6 Redundancy in proprioception at the ankle

There is overwhelming evidence that outputs from specific receptors e.g. muscle spindles are involved in the detection of passive movement and joint position. Robust experimental designs have evolved to establish the subtleties of the underlying physiological processing governing proprioception. Typically, these methodologies have attempted to isolate subtle contributors to awareness of position or movement at the ankle by use of anaesthesia (Konradsen, Ravn et al. 1993; Feuerbach, Grabiner et al. 1994; Vuillerme, Chenu et al. 2006; Down, Waddington et al. 2007; Vuillerme and Cuisinier 2008; Lowrey, Strzalkowski et al. 2010). Anaesthetising the skin has been demonstrated to negatively affect the sense of movement in the hand and fingers (Gandevia and McCloskey 1976; Ferrell and Smith 1988). More recently, Lowrey et al (2010) observed deficits in saggital plane position matching of the ankle after anaesthetising the skin on the dorsum of the foot suggesting that skin receptors play an important role in proprioception at the ankle. It has been previously suggested that the cutaneous input to the sole of the foot from the application of an ankle-foot orthotic can improve acuity in position matching at the ankle (Feuerbach, Grabiner et al. 1994). Conversely, the same group of participants demonstrated no deficit in position matching of the ankle following local anaesthetic injection to the anterior talofibular and / or calcaneofibular ligaments (Feuerbach, Grabiner et al. 1994). Similarly, the administration of a nerve block to anaesthetise the foot and ankle has been demonstrated to have a deleterious effect on the position sense of the ankle (as measured in inversion) (Konradsen, Ravn et al. 1993). However, this effect was negated when participants were requested to actively contract their calf muscles during testing (Konradsen, Ravn et al. 1993). Furthermore, no effect on postural stability during single leg stance was observed as a result of the nerve block (Konradsen, Ravn et al. 1993). This finding suggests that other sources of afferent feedback may either be more relevant in upright tasks, substituted for the lost cutaneous input or both, indicating a redundancy in the proprioceptive system (Konradsen, Ravn et al. 1993). More recent studies have elegantly demonstrated how the central nervous system (CNS) can utilise multiple sources of proprioceptive input to improve the proprioceptive acuity of the ankle by providing biofeedback from the tongue (Vuillerme, Chenu et al. 2006; Vuillerme and Cuisinier 2008). Such augmented feedback is sufficient to overcome the disruptive effects of a vibration stimulus to the Achilles tendon during testing (Vuillerme and Cuisinier 2008). Therefore, the literature suggests that proprioception is not driven by individual peripheral receptors e.g. muscle spindles, joint receptors, ligaments or skin operating in isolation. Rather, a redundancy exists where multiple sources of afferent feedback contribute to ones sense of movement and position.

1.7 Sensation does not equate to perception

It is important to understand the distinction between a sense and a perception. According to Stillman (2002) “sense (sensation) literally means recognising a single specific type of stimulus as, for example, touch or warmth. Perception on the other hand is a cerebral process designed to clarify the nature of the source of a stimulus or stimuli”. A key feature of proprioception is that has been classified as a “perception” (Stillman 2002) which is utilised to inform accurate and coordinated motor control (Sturnieks, Wright et al. 2007). The literature is unequivocal that humans are not consciously aware of every signal generated by proprioceptors (Proske and Gandevia 2012). It would be incorrect to consider proprioception as an internal radar which informs of us of our position or movement at any given point. Rather, a recently proposed model suggests that perception of proprioceptive signal is based upon a central comparison between sensory and motor signals (Bays and Wolpert 2007) as represented schematically in Figure 1.3. According to Proske and Gandevia (2012);

“The process begins with the intention to move (Plan), leading to generation of a motor command and its efference copy/corollary discharge. The Forward model uses the efference copy to calculate the expected outcome, and this is compared with the input (Body in Environment) by means of a difference calculator to quantify the sensory discrepancy that determines what is actually perceived (Perception).”

Therefore, if the predicted sensory feedback matches the actual proprioceptive feedback there should be no actual conscious perception.

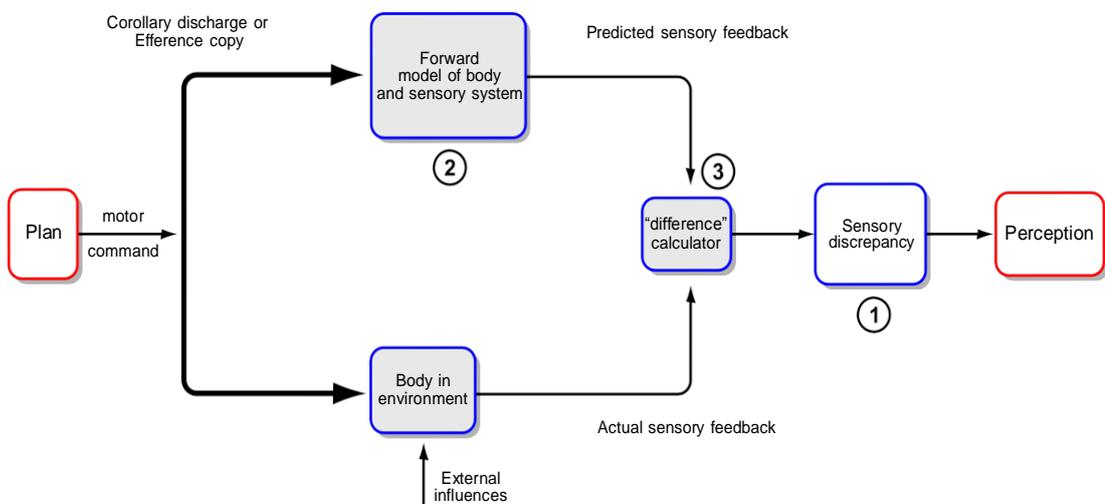


Figure 1.3 Schematic representation of the Forward model of motor control and its sensory cancellation mechanism - reproduced from (Proske and Gandevia 2012)

A key component of this model is that it acknowledges the influence of external factors on the executed movement and thus on any subsequent perception of a difference in performance. However, the majority of proprioception research is based on testing methods which do not incorporate a concept of a changing and unpredictable external environment. Instead, participants are required to perceive changes in their range of motion or joint position such as active repositioning tasks, passive positioning and passive movement detection (Verschueren, Brumagne et al. 2002; Down, Waddington et al. 2007; South and George 2007; Muaidi, Nicholson et al. 2008; Lowrey, Strzalkowski et al. 2010; Salles, Alves et al. 2011). It is true that imposing experimental pain (Matre, Arendt-Neilsen et al. 2002), conditioning contractions (Wise, Gregory et al. 1996; Wise, Gregory et al. 1998), vibrations (McCloskey, Cross et al. 1983) or muscle fatigue (Allen, Leung et al. 2010) and measuring their effect on these tests examines the influence of external factors on the proprioceptive system. However, it may be argued that these methodologies lack ecological validity. In fact, such experimental paradigms have been acknowledged to not be representative of normal function (Proske, Wise et al. 2000). Further highlighting this, a recent review of proprioception literature by two of the most prominent researchers in the area opened with the following statement:

“In everyday activities we depend on signals coming from our moving bodies to be able to respond to the space around us and react rapidly in changing circumstances” (Proske and Gandevia 2012).

In summary, there appears to be a discrepancy in the existing knowledge base between the tests that are currently used to investigate the subtle mechanisms underlying the generation of peripheral proprioceptive signals and normal dynamic function. Current testing methods focus solely on participants' perception of changes in their own configuration. There is currently no described measure of proprioceptive function of the lower limb which considers participants' perception of changes in the environment in which they are operating. Furthermore, testing of the lower limb is largely performed in a seated position which may not reflect to the main functions of the lower limbs – stance and locomotion. As discussed previously, investigation of ankle proprioception has evolved to measurement in standing and the findings of these studies are relevant to the maintenance of upright posture. However, findings from static standing tasks may not be transferrable to more dynamic function of the lower limb such as locomotion. The dynamic function of the lower limbs and its relevance to perception of movement shall be discussed in the following sections.

1.8 The stretch-shortening cycle

As described above, current models of investigating lower limb proprioception are largely based on isolating single joints and assessing participants' acuity in position matching and detection of movement. A critical issue is that it is very difficult to draw inference from such controlled experimental designs to function in the real world where perception of afferent signals may be highly task dependent (Ivanenko, Solopova et al. 2000; Courtine, De Nunzio et al. 2007). The normal dynamic function of the lower limb is to perform repeated stretch-shortening cycles (SSC) (Voigt, Dyhre-Poulsen et al. 1998; Komi and Ishikawa 2006; Ishikawa and Komi 2007) and there is an absence of any testing paradigm that measures participants' proprioceptive function during such tasks.

Traditionally, muscle activity is described in terms of three actions – isometric, concentric and eccentric. However, locomotion would be especially difficult and uneconomical if it was produced by isolated muscle actions. In fact, efficient locomotion is characterised by repeated SSC (Voigt, Dyhre-Poulsen et al. 1998; Komi and Ishikawa 2006; Ishikawa and Komi 2007). In simple terms, a SSC requires the coupling of eccentric and concentric contractions which enhances the concentric phase via neural and mechanical facilitation (Komi 2000). As the foot makes contact with the ground during hopping, the contractile tissues of the gastro-soleus complex are loaded eccentrically. Concurrently, there is an increase in potential energy in the Achilles Tendon as it is primed for elastic recoil (Komi 2000). The rapid increase in muscle length triggers a spinal stretch-reflex contraction which augments the release of the tendon's potential energy (Ishikawa, Komi et al. 2006). This is further supplemented by the neurological drive to perform the task (Ishikawa, Komi et al. 2006). As a result, the torque produced at the ankle joint far exceeds an isolated concentric push off (Komi 2000; Ishikawa, Komi et al. 2006; Ishikawa and Komi 2007; Ishikawa, Pakaslahti et al. 2007). However, it is not the case that we possess one setting for performing repeated SSC and that is applied in all circumstances. Rather, there exists hierarchal control in the central nervous system (CNS) in order to modulate both reflex and muscle activation components of the SSC to optimise safe and efficient performance (Arampatzis, Bruggemann et al. 2001; Leukel, Gollhofer et al. 2008; Taube, Leukel et al. 2012). Such optimisation requires feedforward and feedback processes to adapt in order to ensure a balance between maximal power output and loading injury risk (Taube, Leukel et al. 2012). The pertinent literature of this control system shall be discussed below.

1.8.1 Control from the Spine

Afferent activity from muscle spindles has been linked to the generation of proprioceptive signals from the periphery that are proportionate to the velocity of their stretch (Hall and McCloskey 1983; Refshauge, Chan et al. 1995). During the landing phase of hopping the muscle spindles of gastro-soleus complex become stretched. The rapid change in muscle length drives 1a afferent activity which in turn causes depolarisation of alpha motor neurones in the spinal cord and thus a stretch reflex or short latency response (SLR) is elicited (Bear, Connors et al. 2007). A temporal relationship between the SLR and muscle activation has been demonstrated during drop jumps where the latency of the first peak of muscle activity of the plantarflexors corresponds to the timing of the soleus SLR (Leukel, Taube et al. 2008). Also, the peak EMG signal of the gastrocnemius muscle has been shown to be 2 – 3 times higher when running (a SSC based activity) when compared to the maximum voluntary contraction (no SSC and thus no stretch reflex) (Dietz, Schmidtbleicher et al. 1979). Furthermore, interference of 1a activity via ischemia decreased the gastrocnemius muscle activity peak associated with the SLR during running (Dietz, Schmidtbleicher et al. 1979). Recently, it has been demonstrated that mechanical compression of the calf muscles using a pressure cuff without inducing ischemia reduced the amplitude of the SLR elicited during passive dorsiflexion of the ankle (Leukel, Lundbye-Jensen et al. 2009). Importantly, 1a afferents were implicated as the reduction in SLR was associated with stretch velocity and not stretch amplitude (Leukel, Lundbye-Jensen et al. 2009). When the cuff was applied during hopping the muscle activity associated with the SLR was reduced (Leukel, Lundbye-Jensen et al. 2009). Thus there exists strong evidence that the spinal stretch reflex generates muscular output during a SSC. The critical function of this stretch reflex driven muscle activity during the SSC is to prevent excessive muscle yielding by ensuring that the stiffness of the limb is appropriate to the task (Taube, Leukel et al. 2012). The following section shall explore this concept further.

1.8.2 The Stretch Reflex and Stiffness

Stiffness refers to the resistance to a change in length (Brughelli and Cronin 2008). This concept has been previously investigated in human locomotion (Farley and González 1996; Ferris, Liang et al. 1999) and particularly in hopping tasks (Blickhan 1989; Dalleau, Belli et al. 2004). Locomotion is commonly modelled using a spring mass representation which describes the lower-limb behaving like a spring to progress the centre of mass (Blickhan 1989). The stiffness of the 'leg-spring' is provided by the

articular and periarticular tissues of each joint i.e. hip, knee and ankle. Thus, lower-limb stiffness refers to the stiffness of the entire leg as a system providing resistance to deformation (Brughelli and Cronin 2008). During low load SSC tasks the ankle is considered to be a primary driver of the stiffness of the leg (Farley and Morgenroth 1999) and as such the dynamic stiffness is provided by the gastro-soleus complex and is matched to specific task demands.

Drop jumps have served as a useful model for examining the role of spinal reflexes in the stiffness modulation at the ankle. They require participants to drop from boxes of various heights and to utilise the SSC evoked from landing to rebound into another jump. An important finding from such a research paradigm was that when participants performed drop jumps from 'excessive' heights (80cm) the muscle activity of the soleus at the SLR was significantly lower than for lower drops (30cm) (Komi and Gollhofer 1997). This finding suggested that the CNS may modulate the stretch reflex of the soleus to provide the optimal stiffness level for a given task (Komi and Gollhofer 1997). Subsequently, it was considered that drop landings would not require the high tendomuscular stiffness that provides rebound in drop jumps. Therefore, it was hypothesised that during drop landings participants would exhibit reduced spinal excitability of soleus 1a afferents reflected by reduced H-reflexes (Leukel, Gollhofer et al. 2008). Based on this hypothesis, soleus H-reflex excitability was examined in drop-jumps and drop landings from three different heights (30, 50 and 75 cm). The H-reflex activity was demonstrated to be inversely related to drop height i.e. reduced amplitude was observed as the height of the box increased (Leukel, Gollhofer et al. 2008). A more recent study by the same research group investigated the effect of box height on neuromuscular adaptations obtained from plyometric training using box jumps. Importantly, they demonstrated that observed decreases in H-reflex amplitude with increasing drop height were correlated with reduced ankle stiffness (Taube, Leukel et al. 2012). Thus it is clear that stretch reflex modulation is an effective mechanism to match ankle stiffness for varied loadings during the same task (drop jumps at various heights) and for different tasks (drop jumps vs. drop landings).

In a recent study, Cronin et al (2011) examined the effect of Achilles tendon vibration on SLR amplitude and the yield at the ankle during the contact phase of running at various speeds (7 – 15 km/h). Consistent with previous findings (Leukel, Lundbye-

Jensen et al. 2009), they observed that interference of 1a afferent signalling induced a reduced SLR in the soleus and gastrocnemius muscles at all running speeds tested (Cronin, Carty et al. 2011). This coincided with increased ankle yielding at speeds from 7 to 12 km/h suggesting that the SLR controls the yield of the ankle (and thus the stiffness) after foot contact during the SSC (Cronin, Carty et al. 2011). However, the unique finding was that they observed at the higher speed of 15 km/h the SLR was depressed but the increase in ankle yield was no longer evident suggesting that the SLR has greater importance at slow to moderate running speeds (Cronin, Carty et al. 2011). This is consistent with the assertion that during low load SSC tasks the ankle is a primary driver of the stiffness of the leg (Farley and Morgenroth 1999) whereas for higher loading it may become less relevant (Günther and Blickhan 2002; Hobara, Muraoka et al. 2009).

Thus, the literature suggests that the spinal stretch reflex is gated to modulate stiffness during repeated SSC. It has been proposed that observed reduction in 1a afferent input and muscle activation at the time of the SLR is a protective mechanism that serve to cater for the predicted high load demands of high impact landings (Taube, Leukel et al. 2012). This is based upon the observations of diminished reflex responses in drop landing as compared to drop jumps leading to reduced stiffness and in turn reduced peak stress as muscle function changes from a 'spring' to a 'dampening' unit (Dyhre-Poulsen, Simonsen et al. 1991; Leukel, Gollhofer et al. 2008). Therefore, it appears that the spinal stretch reflex and its modulation are critical to the optimization of stiffness and therefore successful execution of repeated SSC. The following section shall discuss the supraspinal mechanisms contributing to stiffness modulation during the SSC.

1.8.3 Higher Centres

In the above studies utilising drop jumps and drop landings the participants were always aware of the height they were falling (Komi and Gollhofer 1997; Leukel, Gollhofer et al. 2008; Leukel, Taube et al. 2008). Thus, it is reasonable to suggest that they could predict the moment of ground contact and thus the associated stretch of the plantarflexor muscles which may have influenced the muscle activity during the subsequent SSC (Taube, Leukel et al. 2012). McDonagh and Duncan (2002) investigated the interaction of such pre-programmed mechanisms and the stretch reflex using an 'unexpected mechanical event' to distinguish between predictive and reactive control strategies.

Specifically, they examined drop landings where a false floor, collapsed on impact, meaning the subjects descended for an extra 85 ms to a solid floor below (McDonagh and Duncan 2002). When the participants passed through the false floor the amplitude of EMG activity was double that observed when compared to landing directly onto the true floor and occurred in the absence of significant differences in joint rotations between conditions (McDonagh and Duncan 2002). The authors suggested that the increased response observed after passing through the false floor was due to modulation of the spinal stretch reflex via supraspinal centres (McDonagh and Duncan 2002). Finally, they proposed that the response was driven by the absence of the afferent signals that would have been expected immediately after foot contact (McDonagh and Duncan 2002). The suggestion that the increased reflex gain occurred as a result of a mismatch between the expected afferent input and perceived afferent input is consistent with the Forward Model of motor control discussed previously (Bays and Wolpert 2007).

More recently, transcranial magnetic stimulation (TMS) has been utilised to investigate corticospinal activity of participants during drop jumping and hopping to further strengthen the argument implicating the role of higher centres in the control of this dynamic process (Taube, Leukel et al. 2008; Zuur, Lundbye-Jensen et al. 2010). In the case of drop jumps, it has been demonstrated that motor evoked potentials (MEP) from TMS are small in the early contact phase, but become significantly facilitated 120 ms after ground contact (Taube, Leukel et al. 2008). This finding was the direct opposite of H-reflex modulation which was facilitated in the early contact phase (in keeping with the SLR) but declined as the participants approached the push off phase (Taube, Leukel et al. 2008). Thus, the implication is that the early contact phase is largely governed by stretch reflex activity, whereas the later contact phase is influenced by corticospinal mechanisms (Taube, Leukel et al. 2008). In order to directly investigate the role of the motor cortex in this process, low intensity magnetic (below threshold of MEP) was utilised as it has been demonstrated to suppress the EMG of voluntary muscle contraction (Davey, Romaguere et al. 1994). Zuur et al (2010) applied subthreshold magnetic stimulation during hopping and examined its effect on the voluntary muscle activation which occurs on top of the stretch reflex in the early contact phase. When they observed a reduction in the SLR when the stimulation was applied they confirmed that descending drive from the motor cortex directly contributes to the SLR (Zuur, Lundbye-Jensen et al. 2010). Thus, the motor cortex is directly involved in the ground contact phase of SSC activities such as hopping and drop landings.

Clearly, both spinal and cortical mechanisms help to modulate SSC movements. Despite some suggestion of cerebellar involvement in forward planning movement (Bastian 2006; Izawa, Criscimagna-Hemminger et al. 2012), there is no research available which directly implicates specific subcortical brain structures in control of the SSC (Taube, Leukel et al. 2012). Nonetheless, it is evident that during the SSC that multiple processes interact in order to optimise performance. Central to this idea is the interaction of feedforward and feedback control which shall be discussed in the following sections.

1.8.4 Feedforward and feedback control of the SSC

The literature on proprioception describes how 1a afferent fibres from muscle spindles are responsible for a large proportion of sensory information regarding position and movement (Proske and Gandevia 2012). It is evident that such afferent information is critical in the operation of the SSC (Leukel, Taube et al. 2008; Cronin, Carty et al. 2011). Similarly it is evident that peripheral and central mechanisms interact to modulate the execution of SSC (Taube, Leukel et al. 2012) and task specific modulation has been described for both proprioception and the control of the SSC (Dyhre-Poulsen, Simonsen et al. 1991; Ivanenko, Grasso et al. 2000; Courtine, De Nunzio et al. 2007; Leukel, Gollhofer et al. 2008). Furthermore, it has been described how afferent information is modulated so that not all signals are perceived as proprioceptive feedback (Proske and Gandevia 2012). It is evident that we compare our expected sensory feedback to the actual sensory feedback and any discrepancy between the two is perceived as proprioceptive information regarding the performed task (Bays and Wolpert 2007).

Understanding the interaction of feedforward and feedback control systems is fundamental to understanding normal dynamic function. Feedforward control refers to the pre-planned movement which occurs independent of online peripheral feedback (Bastian 2006). By contrast, feedback control utilises online peripheral feedback to either reinforce or refine the current movement (Bastian 2006). Theoretically, the performance of repeated SSC can be classified as a pre-planned movement under feedforward control. This feedforward control is evident in numerous studies investigating the SSC. An example from a previously mentioned study is when participants performed drop jumps from different heights the muscle activity of the soleus at the SLR was significantly lower from 'excessive' heights (80cm) than for lower drops (30cm) (Komi and Gollhofer 1997). This modulation of the

stretch reflex of the soleus to provide the optimal stiffness level for a the specific task requires the CNS to make pre-planned alterations to the behaviour (Komi and Gollhofer 1997). The fast movements involved in SSC tasks cannot accommodate the time lag associated with sensorimotor loops (Wolpert and Flanagan 2010) and so in the case of increasing box height during box jumps the CNS cannot wait to feel the impact level and then choose an appropriate output. Instead such tasks are essentially driven by feed forward control (Taube, Leukel et al. 2012).

A clear example of the feedforward nature of the SSC has been demonstrated via the ‘trampoline aftereffect’ where participants demonstrated measurable changes in motor performance in counter movement jump (CMJ) after exposure to jumping on a trampoline (Marquez, Aguado et al. 2010). The specific biomechanical changes included an increase in leg stiffness and decrease in CMJ height. However, critically the participants misestimated their performance and also reported ‘abnormal subjective sensations’ associated with CMJ performance post trampoline exposure (Marquez, Aguado et al. 2010). It was proposed that the ‘trampoline aftereffect’ may result from errors in the internal model of the participants as a result of the high vertical forces associated with use of the trampoline (Marquez, Aguado et al. 2010). Thus, immediately prior experience can influence the internal model and in turn feedforward control. In a subsequent study, the same research group investigated the neuromuscular after effects in more detail and observed significant increases in the knee extensor muscle activity during the eccentric phase and increased co-activation of the ankle muscles during the concentric phase of CMJ post trampoline use (Marquez, Aguado et al. 2013). They also observed an increase in leg stiffness and a reduction in jump height as per their previous findings (Marquez, Aguado et al. 2010; Marquez, Aguado et al. 2013). They proposed the changes in CMJ observed resulted from a discrepancy between predicted sensory input (efferent copy) and the actual sensory feedback (Marquez, Aguado et al. 2013) which is absolutely consistent with the Forward model of motor control previously described in Figure 1.3 (Bays and Wolpert 2007).

The literature has examined motor performance and the modulation of the SSC in response to external challenges such as changes in surface height or surface integrity. When perceived, these challenges induce a correction to maintain centre of mass (COM) dynamics via feedback control. Thus, the successful performance of repeated SSC involves a complex interaction between both feedforward and feedback control (Taube, Leukel et al. 2012). In fact it has been demonstrated that we can refine and update future performance from a single

miscalculated performance (Marquez, Aguado et al. 2010). In an evolution of the Forward model of motor control, Taube et al (2012) have elegantly illustrated and described this interaction as per Figure 1.4.

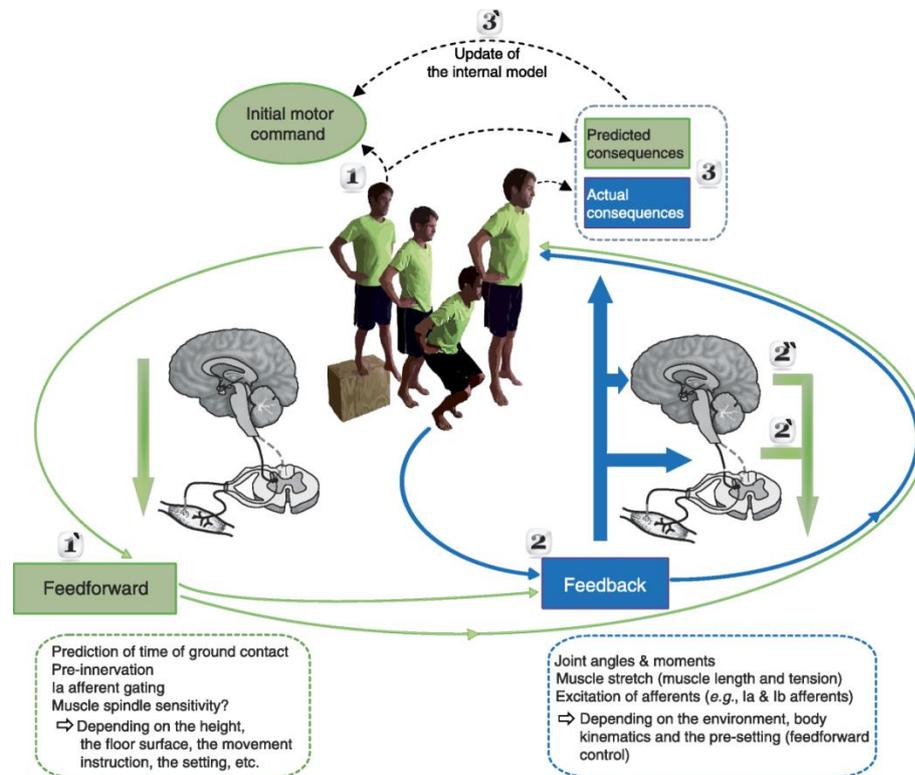


Figure 1.4 A schematic representation of the interaction between feedforward and feedback control during stretch-shortening cycle movement

1) Initial motor command to initiate the movement and to adjust the system in accordance to the expected environmental setting (1'): The feedforward or predictive motor control refers to the portion of the movement that is planned in advance and is not altered by online peripheral feedback. In case of the drop jump, the instant of ground contact can be estimated, and factors like floor surface, aim of the movement (for instance, “to rebound as fast as possible” or “to rebound as high as possible”), and the stability of the environment (e.g., opponent) can be given consideration. Dependent on the situation, the CNS will adjust its activity, like for instance, the amount and duration of preactivation or the Ia afferent gating. 2) At touchdown, peripheral feedback will be generated and can be integrated into the current movement to provide online reinforcement (e.g., activity of the short-latency stretch reflex (2') on top of a supraspinally preprogrammed baseline activity; and/or correction (for instance, if the CNS miscalculated the instant of touchdown or the properties of the landing surface). The feedback loop can involve spinal structures (2') or can be traveling via supraspinal centers (2''). 3) The predicted and the actual consequences of the movement are compared. If they are not in agreement, the internal model has to be updated (3'). This may be the case when the biomechanics of the limb or task have been changed. Consequently, the internal model has to adjust the motor output to the new setting by altering the feedforward command and modifying the gating of afferent integration (e.g., the level of presynaptic inhibition at the spinal level). It has been shown that for the update of the internal model during a series of jumps, information about one single miscalculated jump is sufficient to recalibrate appropriately the internal jump model. Most probably, subjects use the error between the predicted and the actual consequences (sensory feedback) for recalibration. Image and caption reproduced from (Taube, Leukel et al. 2012)

Most studies provide their participants absolute certainty about the task execution i.e. they can visually observe the drop height and predict the moment of ground contact in order to pre-plan the execution of a drop jump. A single study has examined the effect of task uncertainty on the feedforward control of the SSC. Leukel et al (2012) requested participants to perform drop jumps / landings under the following conditions:

1. “No switch”: standard drop jump from a 50cm box
2. “Potentially switch”: Perform a drop jump from a 50cm box, however if an audible cue was heard prior to ground contact participants had to switch from jumping to landing
3. “No switch landing”: Perform a 50cm box landing

They observed that the manner in which participants performed drop jumps during the “Potentially switch” condition was different from the “No switch” condition where there was absolute certainty regarding the planned movement. Specifically, the “Potentially switch” condition resulted in reduced extensor muscle activity at the time of the SLR compared to the “No switch” condition. It was suggested that this performance represented a hybrid movement pattern somewhere between the “No switch” and “No switch landing” conditions (Leukel, Taube et al. 2012). This study clearly demonstrated that the same task (box jump) can be pre-programmed differently as the specific situation requires. An important consideration is that the motor strategy modulation observed may have also reflected the risk of a relatively high impact loading from a 50cm drop. This paradigm has not been tested under lower loading parameters. Therefore, it remains unknown whether the observable differences would be present at lower drop heights or whether the lower limb may possess sufficient mechanical redundancy to negate changes in feedforward control. The interaction of risk, certainty of performance and motor strategy modulation shall be discussed further in the next section.

1.9 Feedforward and feedback control of stiffness - Expected vs. unexpected perturbations

It is clear from the above section that the ‘leg spring’ is not a fixed, innate property of the limb and instead it is adjusted accordingly in anticipation and response to varying task and environmental conditions. In the absence of stiffness modulation, external perturbation would cause a significant compromise of running mechanics, velocity and COM dynamics (Ferris, Liang et al. 1999). For example, rapid adjustment of leg stiffness allows for smooth transition between varying terrains (Ferris, Liang et al. 1999) by offsetting the positional change of the COM caused by a surface change (Moritz, Greene et al. 2004). Thus, the net

disturbance of COM position is optimised to allow task performance and to avoid a catastrophic event such as a fall or abnormal loading which may cause injury. In order to understand the subtleties of stiffness modulation a common research design has emerged based on introducing a variety of external perturbations such as mechanically perturbing the ankle during the swing phase (Scohier, De Jaeger et al. 2012), altering surface stiffness or surface height during SSC activity (Ferris, Liang et al. 1999; Moritz and Farley 2004; van der Linden, Marigold et al. 2007; van der Krogt, de Graaf et al. 2009; van der Linden, Hendricks et al. 2009). Essentially, these external challenges can be categorised as either expected or unexpected, depending on the participants' prior knowledge of the impending perturbation.

The lower limb is capable of feedforward stiffness modulation to cater for expected challenges or perturbations. This has been demonstrated via altered leg kinematics and muscle activity prior to an expected surface change during running (Ferris, Liang et al. 1999). Similarly, participants may adopt an altered box jump strategy to prepare for a potential in-flight switch to a drop landing behaviour if cued to do so (Leukel, Taube et al. 2012). In particular, the effect of a change in surface height on neuromuscular performance during walking has been investigated under both expected and unexpected conditions thus allowing for the distinction between feedforward and feedback control in maintenance of safe COM dynamics (Voigt, Dyhre-Poulsen et al. 1998; van der Linden, Marigold et al. 2007; van der Linden, Hendricks et al. 2009; Masahiro and Shingo 2010). Although there have been multiple different methodologies examining modulation of the leg spring under various challenges, the 'unexpected foot in hole scenario' represents a perfect example of the wider literature. Masahiro and Shingo (2010) instructed healthy participants to walk along a runway where an 8.5cm hole existed above a force plate. They performed two trial types; under one condition they had prior instruction of a potential for a perturbation, and for the other condition there was no instruction regarding the potential for a perturbation. Rapid and large EMG signal responses were detected in both legs when participants unexpectedly encountered an 'unexpected foot in hole scenario'. More specifically, there was an increase in ipsilateral plantarflexor and knee extensor activity and a concurrent increase in the contralateral dorsiflexors and knee flexors (Masahiro and Shingo 2010). This was interpreted as a 'stop walking synergy' i.e. a protective response (Masahiro and Shingo 2010). Furthermore, they observed shorter latencies between the expected heel contact and onset of muscle activity when participants expected a perturbation may occur, demonstrating an anticipatory response (Masahiro and Shingo 2010). In an excellent example of the Forward model of control, it was observed that when the participants did not encounter the

expected normal foot strike, the absence of the expected afferent input triggered muscle responses before the foot actually made contact with the true floor 8.5cm below (Masahiro and Shingo 2010). This finding was consistent with previous findings in walking (van der Linden, Hendricks et al. 2009) and the previously discussed drop jumps with a false floor (McDonagh and Duncan 2002) and drop jump / landing decisions (Leukel, Taube et al. 2012; Scohier, De Jaeger et al. 2012). Finally, they observed that when participants expected that a perturbation may occur, they fundamentally changed their walking strategy by adopting a 'cautious, flat-footed' / decreased dorsiflexed position at foot strike which may be consistent with optimising the soleus response to a potential loss of ground support (van der Linden, Hendricks et al. 2009; Masahiro and Shingo 2010). Such paradigms introduce a perturbation to performance e.g. foot in hole, the participants experience the challenge and the testers measure changes in motor behaviour and kinematics associated with the interaction. The literature has highlighted the involvement of feedforward and feedback mechanisms in this process, thus making proprioception relevant to the modulation of stiffness. However, some gaps exist in the literature. Firstly, it has been suggested that stiffness modulation represents an attempt to offset the net disturbance of COM position and avoid falling, thus allowing task performance to continue safely (Ferris and Farley 1997; Ferris, Liang et al. 1999; Moritz and Farley 2004). Therefore, such paradigms incorporate an element of risk sub-optimal performance (Masahiro and Shingo 2010) or uncertainty in task execution (Leukel, Taube et al. 2012). Considering the 'foot in hole' example, it remains unknown whether unexpectedly encountering smaller perturbation would result in the same motor responses. It is plausible that a smaller hole would represent less risk to task execution but it is unknown whether the same motor responses occur if risk is controlled. There have been no studies to date which examine the thresholds at which a perturbation elicits a motor strategy change. Therefore, it is unknown whether the limb has sufficient mechanical redundancy to accommodate a smaller unexpected 'foot in hole' without a resulting change in motor behaviour. The same considerations may apply to the drop jumps / landings paradigm which introduces task uncertainty in a high load scenario (Leukel, Taube et al. 2012). It is similarly unknown whether the lower limb would have sufficient mechanical redundancy to accommodate task uncertainty without the need to adopt a 'hybrid' motor strategy if the loading parameters (50cm drop) and thus risk were reduced.

A similar paradigm has been applied to hopping where participants have been shown to modify both their mechanics and muscle activity profiles in order to maintain centre of mass dynamics when hopping on a range of surface stiffnesses (Moritz, Greene et al. 2004) and specifically on highly compliant and elastic surfaces (Moritz and Farley 2005). In particular,

when hoppers anticipated an in-flight change in surface stiffness they adjusted their limb kinematics and increased leg muscle pre-activation in order to prepare for the expected change in stiffness demands (Moritz and Farley 2004). In the same cohort, an unexpected change in surface stiffness induced rapid lower limb kinematic changes in order to maintain COM dynamics (Moritz and Farley 2004). However, these changes occur within the latency of the stretch reflex and in the absence of feed-forward anticipatory involvement (Moritz and Farley 2004). Therefore, during this brief period passive dynamics were responsible for the compensatory changes in leg geometry for an unexpected perturbation during hopping (Moritz and Farley 2004; van der Krogt, de Graaf et al. 2009). The literature suggests that different combinations of mechanical and motor control factors interact to maintain optimal and safe loading in the presence of expected and unexpected perturbation during repeated SSC.

In summary, there is a substantial body of research that examines the mechanical changes in the 'leg spring' system under various challenges. The most common experimental paradigm in the literature focuses on the protective feed-forward and feedback responses used to modify leg stiffness to cater for both expected and unexpected external challenges during hopping, drop jumps, drop landings, running and walking. Research thus far has simply classified these perturbations as expected or unexpected. However, the nature of the interaction of the foot and the ground during locomotion is not a simple matter of expected and unexpected perturbation. Proprioception is a key feature of both conditions and thus may play a critical role in stiffness regulation and optimal loading. To date, there are no studies which integrate the concept of proprioception and the modulation of limb stiffness in a changing environment. Specifically, the concept of participants' cognitive perception of external perturbations has not been investigated. Current testing models introduce considerable perturbations that are likely easily perceived by the participants. However, no studies have examined the thresholds at which participants may begin to perceive the perturbation. Furthermore, a distinction has not been drawn between perceived and subliminal perturbations and the subsequent effect on motor performance.

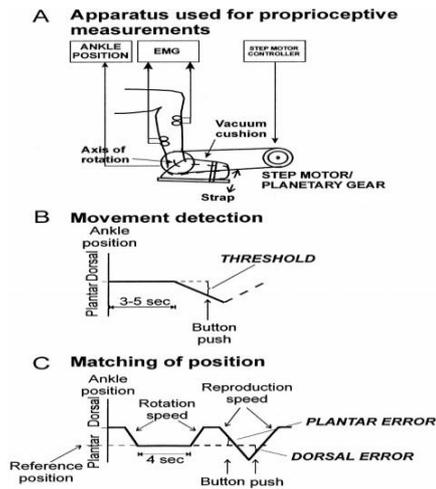
1.10 Merging concepts

Research to date has examined and described in detail many of the process involved in proprioception. This body of knowledge has been dominated by investigation of movement detection and position matching isolated joints in proprioception (Proske and Gandevia 2012). These studies demonstrate that 1a input does not equate to cognitive perception of

movement or position as it is only when a sensory discrepancy exists between the expected and actual sensory feedback perception occurs (Bays and Wolpert 2007). In a more functional context, proprioception testing has evolved to include weight bearing and balance tasks as they pertain to the maintenance of quiet stance (Blaszczyk, Hansen et al. 1993; Blaszczyk, Lowe et al. 1993; Blaszczyk, Lowe et al. 1994; McClenaghan, Williams et al. 1995). However, vibration studies have demonstrated that the gating / utility of proprioceptive information is highly task dependent (Ivanenko, Grasso et al. 2000; Courtine, De Nunzio et al. 2007) making it extremely difficult to draw inference from studies based on open chain isolated movements to more dynamic and functional tasks which can be modelled by repeated SSC. However, to date no studies have investigated the concept of proprioception during repeated SSC activity.

The modulation of the stiffness during the SSC is a complex process which is integral to safe and efficient locomotion (Brughelli and Cronin 2008). It appears that performance of SSC is largely a pre-programmed motor command, however if the expected afferent feedback and the actual afferent feedback don't match then the internal model may be updated to refine future performance and this is perceived cognitively (Taube, Leukel et al. 2012). Thus, proprioception and the modulation of 'leg spring' stiffness appear highly related yet no studies exist which link the concepts of proprioception and motor performance during the SSC. For example, the common research paradigm is to introduce an external perturbation during repeated SSC and to measure the effect on the neuromuscular system as objectified by EMG signals, joint ranges or stiffness measures (Farley and Morgenroth 1999; Ferris, Liang et al. 1999; Arampatzis, Schade et al. 2001; Moritz and Farley 2004; Moritz and Farley 2005). These studies have contributed significantly to our current understanding of lower limb motor control. However, they simply require the participant to endure a perturbation / challenge whilst an external biomechanical or neurophysiological measurement is made. Thus, no insight into the participants' cognitive perception of the perturbation has been reported. At present no studies exist which attempt to merge these concepts and fill the gap in the knowledge base between the highly mechanistic proprioception research and studies examining motor performance of the lower limb in changing environments (See Figure 1.5).

Assessing the physiological mechanisms of proprioception



Stiffness modulation of the leg spring in response to external challenges

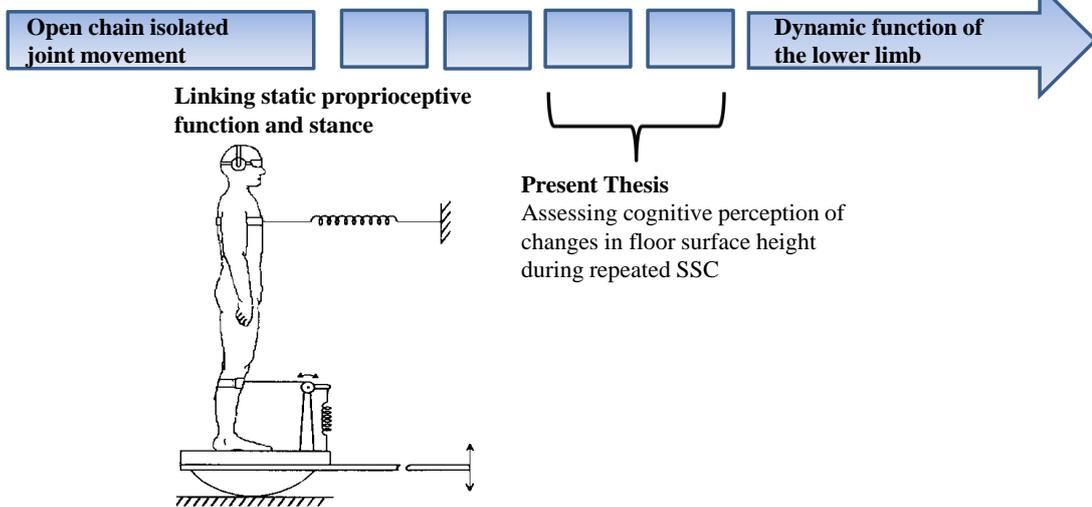
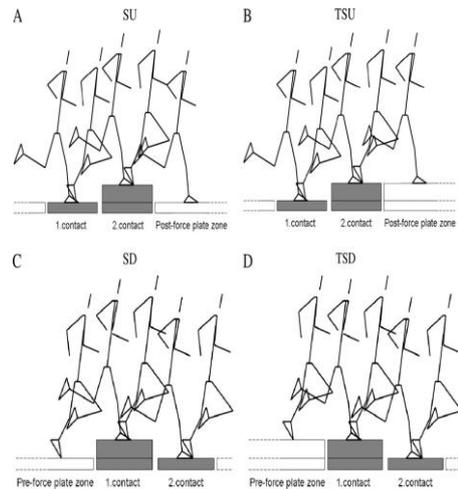


Figure 1.5 A schematic representation of the existing literature - Images reproduced from (Gurfinkel, Ivanenko et al. 1995; Matre, Arendt-Neilsen et al. 2002; Müller and Blickhan 2010)

2. Individual studies and outline of thesis

2.1 Objectives

The following body of work represents a novel shift in the proprioception literature by investigating participants' cognitive perception of changes in floor surface height during repeated SSC. The novelty of the concept created unique challenges as it was not possible to simply utilise pre-existing testing regimes or apparatus. Thus, the first chapters investigate and discuss some of the practical aspects involved in establishing our protocol – the Minimal Perceptible Difference (MPD) test. Specifically, these chapters aim to

1. Investigate the reliability of a sleigh system as an appropriate environment for investigating repeated SSC
2. Introduce and establish the theoretical basis for the MPD test
3. Examine the reliability of the MPD test in a normal population on a within and between day basis

Based upon the above, the methodology established by the MPD test served as a basis to

1. Investigate the effect of expectation of changes in their hopping environment on their subsequent motor strategies
2. Explore the differences in motor strategy response to perceived and unperceived perturbations during hopping
3. Compare the differences between expectation of change in the environment, perception of a change and a subliminal perturbation on participants motor behaviour

2.2 Structure of thesis and outline

Many of the chapters utilise similar methodologies. In order to minimise repetition each aspect of the methodology shall be described in detail upon its first appearance in the thesis and shall be referred back to in subsequent chapters. To facilitate reading, this thesis has been subdivided into:

1. Chapters – sequenced studies examining specific research questions following to the argument and discussion
2. Supplementary Chapters – studies and discussions which expand upon, verify or further substantiate some of the details of the Chapters and are a significant part of this body of work, but would interrupt the reading of the thesis if included in sequence

3. Appendices

The following is an outline of the four Chapters which constitute the body of this thesis:

Chapter 3: Sleigh vs. Upright Hopping

The novel methods of the thesis utilises a sleigh hopping model and this chapter justifies the use of the sleigh and describes its use. Furthermore, this chapter investigated whether one is examining the same domain in upright and sleigh hopping. The primary research question of this study was to examine differences between leg stiffness and the kinematic profile of the ankle during self-paced hopping on the sleigh and self-paced upright hopping. This was achieved by answering the following research questions:

1. Are ankle kinematic variables and leg stiffness as reliable during self-paced hopping on the sleigh as during upright hopping on a within and between day basis?
2. Are upright hopping and sleigh hopping the same task?
 - Are ankle kinematic variables and leg stiffness significantly different between self-paced upright and sleigh hopping?
3. Are we measuring the same domain?
 - Do leg stiffness and kinematic variables of the ankle during the contact phase correlate when examined between self-paced upright and sleigh hopping?

This chapter parallels and is supported by other studies included in the thesis as:

- Supplementary Chapter 1 - “Estimates of leg stiffness during low-load plyometrics” (Grisbrook et al Submitted Manuscript)
- Appendix 1 - “A Novel Sledge-Jump System that is Reliable for Measuring the Motor Correlates of the Stretch-Shortening Cycle” (Debenham, Travers et al, Submitted Manuscript)

Chapter 4: Stability of lower limb minimal perceptible difference in floor height during hopping stretch-shortening cycles (Travers, Debenham et al. 2013)

Chapter 4 introduces and establishes the theoretical basis for the MPD test. It aims to compare participants' ability to cognitively detect changes in floor surface height during repeated SSCs across different hopping strategies. A secondary aim is to establish the reliability of the novel MPD test on a within and between day basis.

Chapter 5: Motor strategy modulation when expecting an external perturbation – a protective motor response or a searching sensory strategy?

This chapter utilises the protocol established by the MPD test to investigate the effect of expectation of a change in floor height on motor strategy during sleigh hopping. The authors are unaware of any studies which purely describe the role of expectation of a safe, subtle and familiar change in the foot / floor interface on the mechanics and neural drive during hopping. The following research questions are addressed:

1. Does expectation of a change in surface height during bilateral hopping alter the sagittal kinematic profile of the ankle?
2. Does expectation of a change in surface height during bilateral hopping alter the EMG profile of the ankle?

Chapter 6: Does perception of an expected perturbation induce a motor strategy response?

Chapter 7 examines whether the same performance changes occur for a subliminal perturbation as for one that is above the threshold of cognitive perception. This novel approach merges the concepts of perception and modulation of stiffness and motor behaviour. The following question is addressed:

- Does perception of a change in surface height during bilateral hopping alter the sagittal kinematic and muscle activity profile of the perturbed ankle?

Chapter 7: Perceived changes in floor height occur without detectable changes in range

Chapter 7 further explores whether the same performance changes occur for a subliminal perturbation as for one that is above the threshold of cognitive perception. The following research questions are addressed:

- Does the default sagittal plane kinematic profile of the ankle change for perceived and subliminal changes in floor surface height during the MPD test?
- Does the default muscle activity profile of the ankle change for perceived and subliminal changes in floor surface height during the MPD test?

A Note on Chapters 5, 6 & 7

Chapters 5, 6 and 7 are based on the same data collection, however the data presented in each of the different analyses represents different epochs during participants performance of the testing protocol.

Chapter 8: Discussion and summary of major findings

Chapter 9: Discussion and summary of major findings

3. Upright Hopping vs Sleigh Hopping

3.1 Introduction

Safe and efficient locomotion is characterised by repeated stretch-shortening cycles (SSC) (Voigt, Dyhre-Poulsen et al. 1998; Komi 2000; Ishikawa and Komi 2007). This has been represented using a spring mass model which describes the lower-limb behaving like a spring to progress the centre of mass (Blickhan 1989; Le Meur, Dorel et al. 2012). Leg stiffness during locomotion is provided by articular and periarticular tissues and is modulated to reflect the given task demands (Yen, Auyang et al. 2009; Yeadon, King et al. 2010) via complex interaction with the peripheral and central nervous systems (Santello 2005). Understanding the complex mechanical behaviour of the limb during normal function has become topical for clinicians and researchers engaged in human performance, injury management and rehabilitation (Butler, Crowell et al. 2003; Zifchock, Davis et al. 2006; Bryant, Newton et al. 2009; Yen, Auyang et al. 2009; Zeni Jr and Higginson 2009; Hobara, Inoue et al. 2010; Hobara, Kato et al. 2012).

A common experimental model has emerged using repeated hopping trials to investigate stiffness during the SSC (Farley, Blickhan et al. 1987; Cavagna, Franzetti et al. 1988; Blickhan 1989; Ferris and Farley 1997; Farley and Morgenroth 1999; Dalleau, Belli et al. 2004; Allison, Utsunoniya et al. 2005; Bobbert and Richard 2011). However, multiple hopping trials present challenges as data may be confounded by factors such as balance and fatigue, and thus researchers have developed various 'sleigh' systems to unload the body mass and facilitate collection of data from multiple hops / SSC (Kramer, Ritzmann et al. 2010; Merritt, Raburn et al. 2012; Furlong and Harrison 2013). In order to confidently rely on data derived from such sleigh systems it is necessary to quantify their reliability for pertinent derived variables (Joseph, Bradshaw et al. 2013). In particular, the within-day and between-day reliability of lower limb kinematics and biomechanical measures of the SSC during self-paced hopping on a custom built sleigh apparatus has been investigated in a healthy population (Debenham, Travers et al., Submitted Manuscript - see Appendix 1). Debenham, Travers et al (Unpublished, see Appendix 1) reported strong reliability for lower limb stiffness and kinematic variables of the ankle joint during self-paced sleigh hopping. This is important as the ankle has been reported to be the critical joint for stiffness modulation of the leg during low load tasks such as submaximal hopping (Farley and Morgenroth 1999; Moritz, Greene et al. 2004). Debenham, Travers et al (Submitted Manuscript - see Appendix 1) have further justified the use of the sleigh hopping model for examining performance at the ankle joint by allaying concerns regarding instability in

performance due to the unfamiliarity of sleigh hopping or the impact of factors such as fatigue and balance.

Sleigh hopping is a form of unloaded hopping which remains invalidated against upright hopping. Despite strong reliability of leg stiffness and ankle kinematic variables during sleigh hopping it remains unclear if the function of the ankle joint during sleigh hopping resembles that for upright hopping. Therefore, in order to draw clinical inference from data regarding performance of the ankle during sleigh hopping it is necessary to investigate whether we are examining the same domain in upright and sleigh hopping. In essence, one cannot assume that sleigh hopping is reflective of upright hopping without comparing the two tasks. To this end, the primary research question of this study was to investigate whether leg stiffness and the kinematic profile of the ankle during self-paced hopping on the sleigh resemble self-paced upright hopping. This was subdivided as follows:

4. Are ankle kinematic variables and leg stiffness as reliable during self-paced hopping on the sleigh (SH) as during upright hopping (UH) on a within and between day basis?
5. Are UH and SH the same task?
 - Are ankle kinematic variables and leg stiffness significantly different between self-paced upright and sleigh hopping?
 - Do leg stiffness and kinematic variables of the ankle during the contact phase correlate when examined between self-paced upright and sleigh hopping?

3.2 Methods

This study utilised a repeated measures within-subject experimental design. Participants attended two testing sessions spaced one week apart at the Motion Analysis Laboratory, School of Physiotherapy and Exercise Science, Curtin University. Thirteen healthy participants (7 females and 6 males; mean age 27 years, height 170.3 cm, mass 69.6 kg) were included in this study. Inclusion criteria required that they were free of any pain or functional limitations. Ethical approval was granted by the Curtin University Human Research Ethics Committee (PT0145/2009), and informed consent was obtained for all participants prior to testing.

3.2.1 The Sleigh

This study utilised a custom built sleigh apparatus as pictured in Figure 3.1. The sleigh was instrumented with an AMTI force plate sampling at 1000Hz as a landing platform allowing establishment of critical event markers during hopping e.g. contact and flight phases. The force plate was placed orthogonally to the axis of the sleigh (20 degrees) and attached to a rigid steel structure.

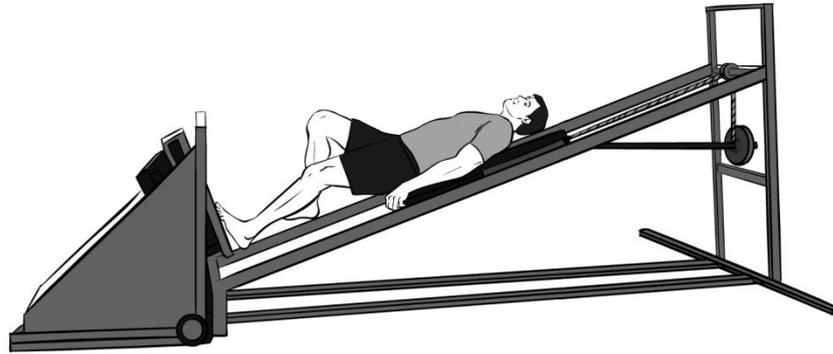


Figure 3.1 Hopping on the custom built sleigh apparatus.

Image reproduced from (Gibson, Campbell et al. 2013)

3.2.2 Procedure

On each testing occasion the participants were requested to perform ten trials of continuous sub-maximal hopping under two different conditions:

1. Upright Hopping (UH): required the participants to hop whilst standing upright at their self-selected frequency on their dominant leg on an AMTI force plate sampling at 1000Hz. This force plate was embedded into the floor of the gait / motion analysis laboratory. They were instructed to hop at a sub-maximal level, described as an effort they could maintain 'indefinitely'.

2. Sleigh Hopping (SH): participants kept the opposite limb in a flexed position, resting supported on the sleigh and held onto the sliding tray of the sleigh in order to stabilise the thorax and upper limbs (See figure 3.1). They were instructed to hop using their non-dominant leg whilst minimising any associated knee flexion (no external fixation was used), thus driving hopping from their ankle. They were instructed to hop at a sub-maximal level, described as an effort they could maintain ‘indefinitely’.

Each trial lasted thirty seconds and was separated with a ninety second rest period between trials. The testing procedure outlined above was repeated one week later. Prior to testing, participants were given a 5 minute familiarisation period in order to become accustomed to hopping on the sleigh. To determine leg dominance the participants nominated the foot they would normally use to kick a ball. This was considered the dominant leg (Witvrouw, Danneels et al. 2003).

3.2.3 3-D Motion Analysis

Fourteen Vicon (Oxford metrics, Inc.) infra-red cameras, sampling at 250Hz, captured the movement of reflective markers that were applied to the lower limbs in accordance with a cluster based model (Besier, Sturnieks et al. 2003). Reflective markers were attached to anatomical landmarks around the pelvis, thigh, leg and foot (anterior superior iliac spine (ASIS), iliac crest, iliotibial band (ITB), tibial shaft, calcaneus, and 1st and 5th metatarsal heads. A validated anatomical marker set (with reconstruction errors of < 1 mm (Ehara, Fujimoto et al. 1995; Richards 1999)) and model (Besier, Sturnieks et al. 2003) was used to generate a 3D, anatomically relevant, reconstruction of the lower limb, including a foot, leg, and thigh segment (See figure 3.2). The validated mathematical model (International Society of Biomechanics, (Wu, Siegler et al. 2002) following ZXY order of rotations (Grood and Suntay 1983) generated sagittal plane ankle range of motion.

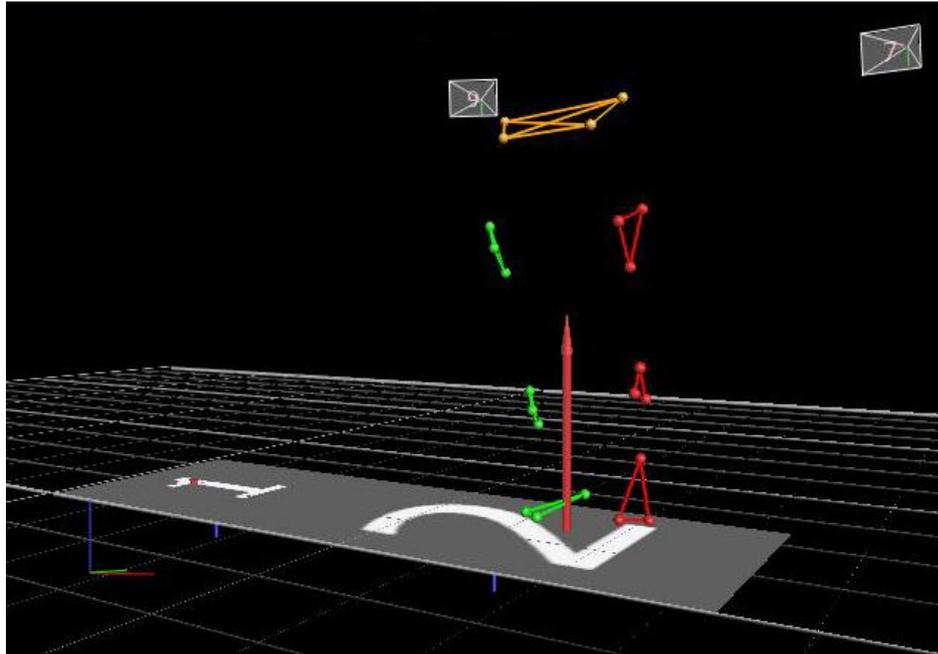


Figure 3.2. Reconstruction of the lower limbs during upright hopping on the right leg using Vicon Nexus 3-D Motion Analysis System

Note: Yellow markers represent the pelvis with markers placed upon the ASIS and iliac crests; Green markers represent the right (hopping) leg with a cluster of 3 markers over the ITB approximately mid shaft of the femur, a cluster of 3 markers over the mid-portion of the tibial shaft and individual markers overlying the calcaneus, and 1st and 5th metatarsal heads; Red marker represent the left leg; The red arrow represents the ground reaction vector from the force plate (Labelled 2); Labels 7 and 9 represent two infra-red cameras which were part of a cluster of 14 cameras.

3.2.4 Kinematic Variables

A customised LabVIEW program (National Instruments Corporation) was utilised to output the following variables in order to develop a profile of sagittal plane ankle excursion during the contact phase for SH and UH:

1. Ankle angle at contact in sagittal plane (θ_{contact})
2. Peak dorsiflexion angle (θ_{peak})
3. Stretch Amplitude ($\theta_{\text{c-p}}$): the change in the ankle joint angle from landing (contact) to the most dorsiflexed point
4. Ankle angle at take-off θ in sagittal plane ($\theta_{\text{take-off}}$)

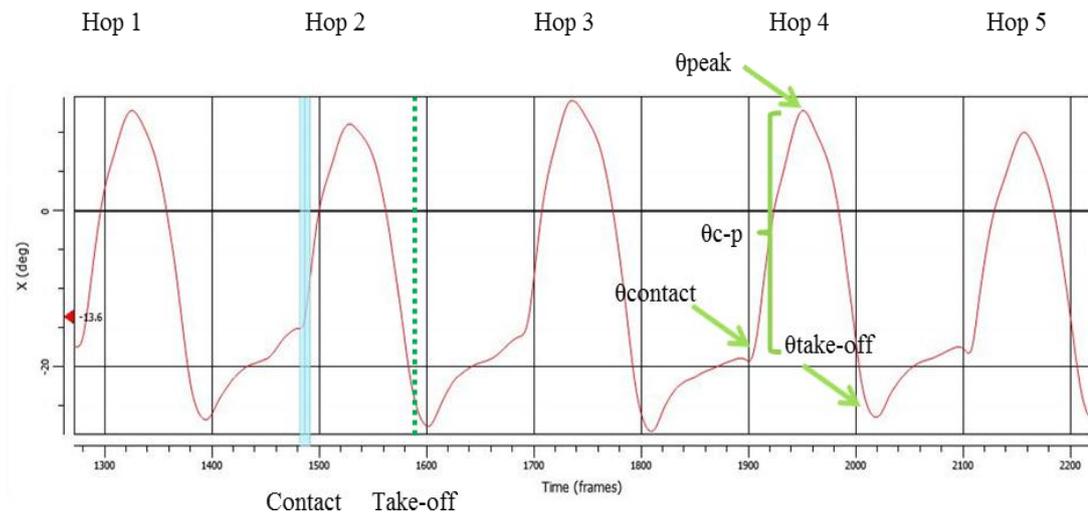


Figure 3.3 Sagittal plane ankle angle trace from Vicon Nexus System (Oxford metrics, Inc.) during 5 consecutive hops on the sleigh.

Note: the vertical axis displays the ankle angle in the X plane / saggital plane; the horizontal axis represents time as measured by frames (1000Hz); minus values represent plantarflexion; The limits of the contact phase are indicated for hop 2 via contact (solid blue line) and take-off (dotted green line); The kinematic variables are indicated (green arrows) for hop 4.

3.2.5 Leg Stiffness

Leg stiffness (K) was estimated using a spring-mass model (Blickhan 1989). Figure 3.4 shows the formula (Dalleau, Belli et al. 2004) that was used to calculate K with the flight (t_f) and contact times (t_c) determined from force plate contact and toe off data. Of note, the conventional unit for this measure is $\text{kN}\cdot\text{m}^{-1}$ and it is a novel approach to estimate leg stiffness during sleigh hopping. The validity of this measure and its units has been demonstrated on the present sleigh system (See Box 1 and Supplemental Chapter 1 – Estimates of Leg Stiffness during low-load plyometrics).

$$K = \frac{(M \times \Pi(t_f + t_c))}{(t_c^2 ((t_f + t_c / \Pi) - (t_c / 4)))}$$

Unit: $\text{kN}\cdot\text{m}^{-1}$

Figure 3.4 Formula for estimating leg stiffness

Stiffness = K; M = total body mass; t_f = flight time; t_c = ground contact time

3.2.6 Analysis

As recommended by Debenham, Travers et al (Submitted Manuscript - see Appendix 1) reliability analysis focused on the first 30 hops of each trial and trial data was pooled for trials 2 – 10 for each testing occasion. Analysis was performed using statistical software for analysis (IBM SPSS Statistics version 2: IBM Corp©, Chicago, IL, USA). Intra-class Correlation Coefficients (ICC) with 95% Confidence Intervals (CI) were derived to examine the reliability of lower limb stiffness, θ_{contact} , θ_{peak} , $\theta_{\text{c-p}}$, $\theta_{\text{take-off}}$ for test day 1 and test day 2. ICC values above 0.70 were considered to represent strong reliability (Portney and Watkins 2009). A paired samples t-test was performed on pooled data from trials 2 -10 from test day 1 for lower limb stiffness, θ_{contact} , θ_{peak} , $\theta_{\text{c-p}}$, $\theta_{\text{take-off}}$ to determine any significant difference between UH and SH. Pearson correlation was calculated for the same pooled data and values from 0.5 to .75 were considered to represent moderate to good correlation and values above .75 were considered to represent strong to excellent correlation (Portney and Watkins 2009). An alpha of 0.05 was used to represent statistical significant for all comparisons. For kinematic analysis there was a loss of one set of data due to a marker set failure.

3.3 Results

3.3.1 Within –Day Reliability

The participants demonstrated excellent reliability for θ_{contact} , θ_{peak} , $\theta_{\text{c-p}}$, $\theta_{\text{take-off}}$ and stiffness for test days 1 and 2 for both upright and sleigh hopping. This is represented in Tables 3.1 and 3.2 and graphically in Figures 3.5 and 3.6 below.

Table 3.1 Day1 reliability of Kinematic Variables (in deg) and Stiffness (kN.m-1) comparing upright and sleigh hopping

Variable	Condition			95% CI		SEM	95% CI		
		Mean	SD	Lower	Upper		ICC	Lower	Upper
θ_{contact}	Upright	-9.4	8.9	-10.6	-8.2	0.6	0.995*	0.989	0.998
	Sleigh	-21.7	7.6	-22.9	-20.4	0.6	0.993*	0.984	0.998
θ_{peak}	Upright	16.0	2.8	15.4	16.5	0.3	0.991*	0.980	0.997
	Sleigh	1.4	5.8	0.6	2.2	0.4	0.995*	0.989	0.998
$\theta_{\text{c-p}}$	Upright	25.4	8.7	24.3	26.4	0.6	0.996*	0.992	0.999
	Sleigh	23.1	7.1	21.8	24.4	0.7	0.991*	0.980	0.997
$\theta_{\text{take-off}}$	Upright	-14.8	9.1	-17.2	-12.3	1.3	0.981*	0.959	0.993
	Sleigh	-30.1	8.1	-32.0	-28.3	0.9	0.987*	0.973	0.996
Stiffness	Upright	15.3	5.0	14.8	15.7	0.2	0.998*	0.995	0.999
	Sleigh	8.5	3.3	8.1	8.9	0.2	0.996*	0.992	0.999

Note: SD = Standard Deviation; CI = Confidence Interval; SEM = Standard Error of Measurement; ICC = Intraclass Correlation Coefficient; * indicates $p < 0.001$; Negative values refer to plantarflexion; All angles in degrees; Stiffness measured in kN.m^{-1}

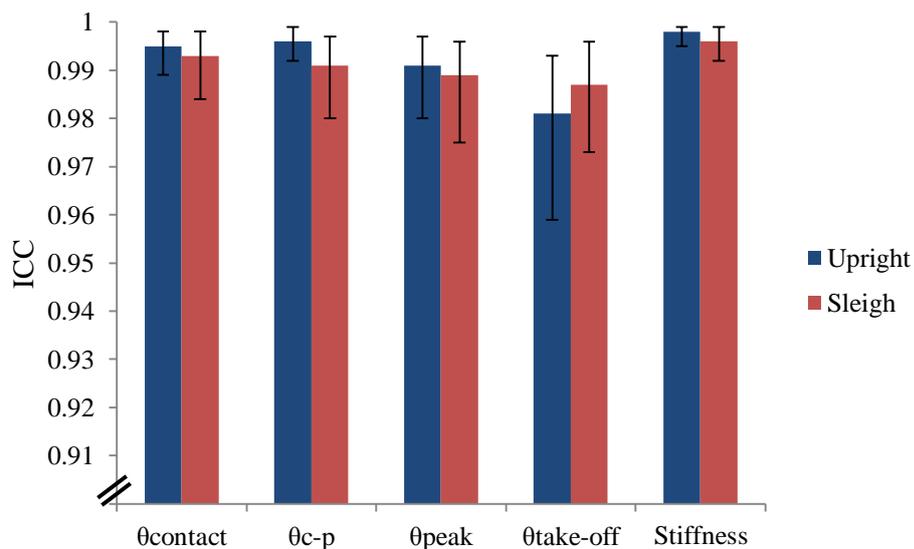


Figure 3.5 Day1 Intraclass Correlation Coefficients (ICC) and 95% Confidence Intervals (error bars) for Kinematic Variables and Stiffness for upright and sleigh hopping

Table 3.2 Day2 reliability of Kinematic Variables (in deg) and Stiffness (kN.m-1) comparing upright and sleigh hopping

Variable	Condition			95% CI		SEM	ICC	95% CI	
		Mean	SD	Lower	Upper			Lower	Upper
θ_{contact}	Upright	-8.2	9.2	-9.2	-7.2	0.5	0.997*	0.994	0.999
	Sleigh	-20.7	9.4	-22.4	-19.1	0.8	0.992*	0.983	0.998
θ_{peak}	Upright	14.0	3.7	12.8	15.1	0.6	0.975*	0.945	0.992
	Sleigh	-2.0	6.2	-3.0	-0.9	0.6	0.992*	0.983	0.998
$\theta_{\text{c-p}}$	Upright	22.2	7.3	21.3	23.1	0.5	0.996*	0.992	0.999
	Sleigh	20.4	7.6	19.0	21.8	0.7	0.991*	0.981	0.997
$\theta_{\text{take-off}}$	Upright	-12.2	11.4	-13.8	-10.6	0.8	0.995*	0.996	0.995
	Sleigh	-29.2	8.5	-30.5	-27.9	0.7	0.994*	0.987	0.998
Stiffness	Upright	17.1	6.4	16.7	17.5	0.2	0.999*	0.998	0.999
	Sleigh	9.3	4.3	8.7	10.0	0.3	0.994*	0.988	0.998

Note: SD = Standard Deviation; CI = Confidence Interval; SEM = Standard Error of Measurement; ICC = Intraclass Correlation Coefficient; * indicates $p < 0.001$; Negative values refer to plantarflexion; All angles in degrees; Stiffness measured in kN.m^{-1}

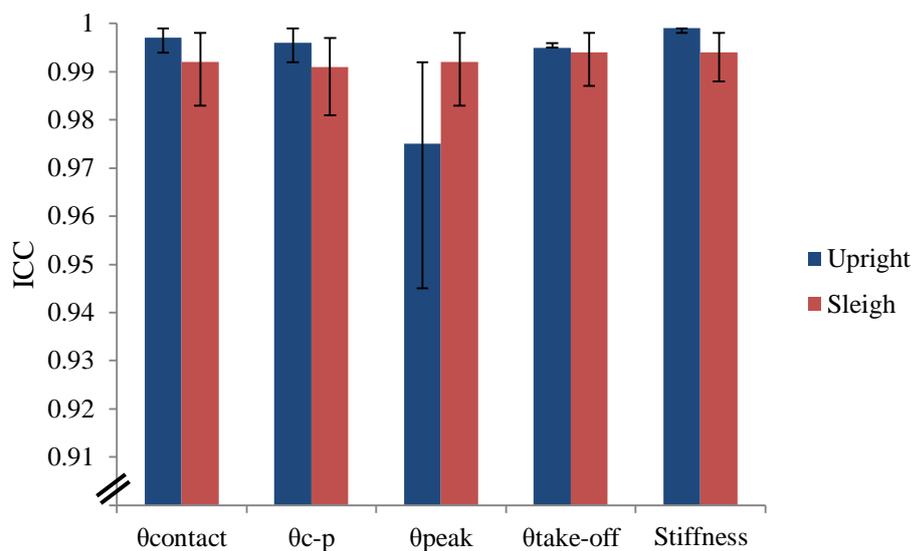


Figure 3.6 Day2 Intraclass Correlation Coefficients (ICC) and 95% Confidence Intervals (error bars) for Kinematic Variables and Stiffness for upright and sleigh hopping

3.3.2 Between –Day Reliability

The participants demonstrated strong to excellent between-day reliability for θ_{contact} , $\theta_{\text{c-p}}$, $\theta_{\text{take-off}}$ and stiffness for both sleigh and upright. θ_{peak} had strong between-day reliability for sleigh hopping but was much less reliable for upright hopping. This is represented in Table 3.3 and graphically in Figure 3.7.

Table 3.3 Between-day reliability of Kinematic Variables (deg) and Stiffness (kN.m-1) comparing upright and sleigh hopping

Variable	Condition	Pooled		95% CI		SEM	ICC	95% CI	
		Mean	SD	Lower	Upper			Lower	Upper
θ_{contact}	Upright	-8.8	9.0	-14.3	-3.3	2.8	0.904*	0.702	0.975
	Sleigh	-21.2	8.5	-30.0	-12.4	4.5	0.717	0.017	0.919
θ_{peak}	Upright	15.0	3.3	10.4	19.5	2.3	0.493	-0.762	0.854
	Sleigh	-0.3	6.0	-6.2	5.6	3.0	0.746	0.117	0.927
$\theta_{\text{c-p}}$	Upright	23.8	8.0	19.3	28.2	2.3	0.921*	0.726	0.977
	Sleigh	21.7	7.4	16.3	27.2	2.8	0.856	0.500	0.959
$\theta_{\text{take-off}}$	Upright	-13.5	10.2	-21.2	-5.8	3.9	0.852*	0.516	0.955
	Sleigh	-29.7	8.3	-38.2	-21.1	4.4	0.723*	0.093	0.916
Stiffness	Upright	16.2	5.7	13.0	19.4	1.7	0.915*	0.721	0.974
	Sleigh	8.9	3.8	7.0	10.9	1.0	0.929*	0.769	0.978

Note: SD = Standard Deviation; CI = Confidence Interval; SEM = Standard Error of Measurement; ICC = Intraclass Correlation Coefficient; * indicates $p < 0.001$; Negative values refer to plantarflexion; All angles in degrees; Stiffness measured in kN.m-1

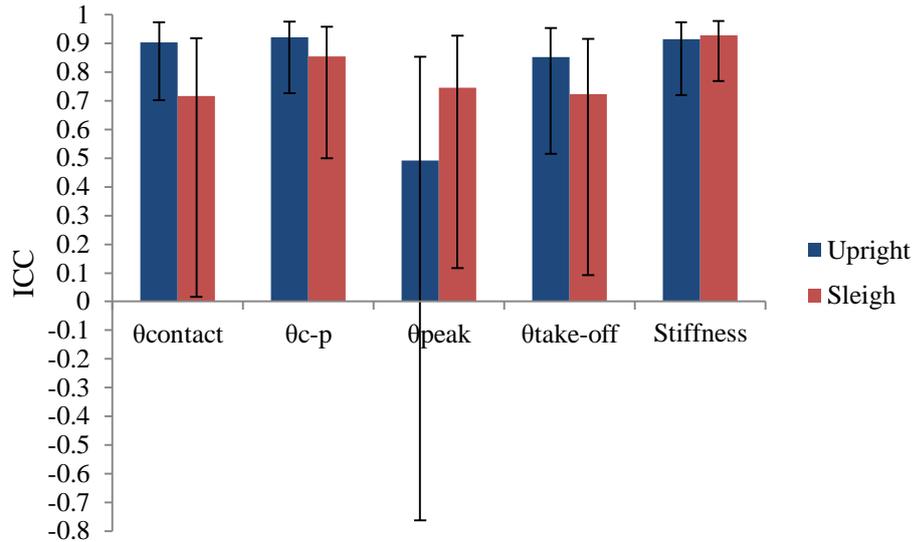


Figure 3.7 Between-day Intraclass Correlation Coefficients (ICC) and 95% Confidence Intervals (error bars) for Kinematic Variables and Stiffness for upright and sleigh hopping

3.3.3 Are Upright Hopping and Sleigh Hopping the same task?

The θ_{contact} was 9.41 degrees of plantarflexion for UH and 21.68 degrees of plantarflexion for SH and this difference was statistically significant ($p < .001$). During the contact phase the θ_{peak} was 15.93 degrees dorsiflexion for UH and 1.41 degrees dorsiflexion for SH, this also represents a significant difference between conditions ($p < .001$). However, there was not a significant difference between $\theta_{\text{c-p}}$ for either condition ($p = .147$) with participants demonstrating a stretch amplitude of 25.35 degrees for UH and 23.09 degrees for SH. The contact phase finished with the ankle in 14.75 degrees plantarflexion for $\theta_{\text{take-off}}$ during UH and 30.14 degrees for SH ($p < .001$). Expectedly, stiffness was statistically significantly ($p < .001$) higher for UH (15.27 kN.m^{-1}) than for SH (8.52 kN.m^{-1}). These results confirm that the operation of the ankle is significantly different between UH and SH. Below, table 4 outlines the differences between and correlation of kinematic variables and lower limb stiffness between UH and SH for test day 1.

Table 3.4 Differences and Correlation between Kinematic Variables (in deg) and Stiffness (kN.m-1) between upright and sleigh hopping

Variable	Condition	Mean	SD	P value	Pearson
					Correlation
θ_{contact}	Upright	-9.4	8.9	<.001*	.610*
	Sleigh	-21.7	7.6		
θ_{peak}	Upright	15.9	2.8	<.001*	0.364
	Sleigh	1.4	5.8		
$\theta_{\text{c-p}}$	Upright	25.4	8.7	.147	.817*
	Sleigh	23.1	7.1		
$\theta_{\text{take-off}}$	Upright	-14.8	9.1	<.001*	0.514
	Sleigh	-30.1	8.1		
Stiffness	Upright	15.3	5.0	<.001*	.887*
	Sleigh	8.5	3.3		

Note: SD = Standard Deviation; * indicates $p < 0.001$; Negative values refer to plantarflexion; All angles in degrees; Stiffness measured in kN.m-1

3.4 Discussion

3.4.1 Reliability

This study utilised a within-subject repeated measures design to examine leg stiffness and the kinematic profile of the ankle during self-paced, sub-maximal hopping under two conditions - SH and UH. Leg stiffness, θ_{contact} , θ_{peak} , $\theta_{\text{c-p}}$ and $\theta_{\text{take-off}}$ demonstrated excellent reliability on both test occasions (all ICC > .975). The hypothesis that the kinematic profile of the ankle is reliable during both UH and SH has been supported for within-day testing. Debenham, Travers et al (Unpublished, See Appendix 1) demonstrated moderate to strong temporal reliability for hopping performance for sleigh hopping. We add to these findings by comparing the relative reliability variables in both standing and sleigh self-paced hopping. The within-day reliability ICC observed in the present are extremely

high (all $>.982$), however this is comparable to ICC reported by other researchers examining the reliability of peak force (ICC = $.984$), flight time (ICC = $.987$) during consecutive SSC on a sleigh system (Flanagan and Harrison 2007). Similarly, the within trial ICC for flight time and contact time on an adapted sleigh system isolating the ankle complex has been reported to be in the range of $.992$ to $.999$ (Furlong and Harrison 2013). Importantly, flight time and contact time are the primary variables used for the present calculation of leg stiffness (Dalleau, Belli et al. 2004) and the ICC values observed for leg stiffness are similarly high. The consistency of reliability analysis between the current research and previous work gives confidence in reporting such strong reliability for leg stiffness for both upright and sleigh hopping.

The present findings add to the knowledge base by also reporting within and between day kinematic data specific to the ankle joint. On a between-day basis all variables for SH demonstrated strong reliability as determined by our 0.7 cut-off as outlined in our methods (Portney and Watkins 2009). Leg stiffness, θ_{contact} , $\theta_{\text{c-p}}$ and $\theta_{\text{take-off}}$ demonstrated strong between-day reliability for UH (all ICC $>.852$). However, θ_{peak} had poor between day reliability (ICC = $.493$; 95% CI range $-.762$ to $.854$). The hypothesis that the presented derived variables are reliable during both UH and SH has been supported on a between-day basis, with the exception of θ_{peak} .

As previously mentioned, equally strong reliability of similar derived variables have been demonstrated on a comparable sleigh system (Furlong and Harrison 2013). However, an important methodological difference exists in the present study where the participants were specifically requested to hop at an effort and rate they could maintain ‘indefinitely’ in order to yield a self-selected hopping frequency and thus an unconstrained flight time, contact time and leg stiffness. Conversely, other studies have constrained both flight and contact times instructing their participants to perform with maximal leg stiffness in order to yield a more reliable performance (Kramer, Ritzmann et al. 2010; Furlong and Harrison 2013). Therefore, it is particularly noteworthy that the reliability scores are so high on both a within and between day basis in the present work. The distinction may be drawn between a reliable hopping performance by constraining leg stiffness and a reliable hopping performance in a low stiffness environment (sleigh) with constrained movement, this concept shall be explored in the next sections.

3.4.2 Are Upright Hopping and Sleigh Hopping the same task?

The results confirm that the operation of the ankle is significantly different for self-paced between UH and SH. Therefore, despite the similar θ_{c-p} upright hopping and sleigh hopping are clearly not the same task. This differs from previous findings where the ankle kinematic profile was observed to be similar between upright and sleigh performances (Kramer, Ritzmann et al. 2010). However, a critical difference in the methodologies was that Kramer et al (2010) were examining reactive jumps whereby the knees and hips contributed significantly to the production of force and thus were operating at very different loading parameters. Effort was made in the present study to localise the movement to the ankle joint for SH by explicitly instructing participants to resist active knee flexion and to drive the hopping performance from the ankle. It has been demonstrated experimentally that such instruction leads to a SH performance where stiffness is largely attributable to the ankle in sub-maximal sleigh hopping (See Box 1).

Box 1. Measuring ankle stiffness using a leg stiffness estimate

Leg stiffness is a key derived variable of this body of work and has been calculated using a field based measurement of leg stiffness (K_D) where stiffness is calculated using body mass, contact time and flight time measured from the force plate (Dalleau, Belli et al. 2004). This method has been validated against the gold standard method (K_C) (Cavagna, Franzetti et al. 1988) and they were found to be highly correlated in submaximal upright hopping ($r=0.94$) for healthy adults. However, this method has not been validated for sleigh hopping. Furthermore, the methodology of this thesis required participants to hop on the sleigh apparatus with a straight knee and drive the performance with the ankle. Theoretically, making the ankle joint complex the primary driver of leg stiffness would mean that the leg stiffness estimate would indirectly measure ankle stiffness (K_{ankle}). Once again, no other authors had compared the leg stiffness estimate to the gold standard measurement of joint stiffness (Farley and Morgenroth 1999) during sleigh hopping. Therefore, it was necessary to determine if

1. The K_D estimate is valid for sleigh hopping and
2. whether K_D is actually a measure of K_{ankle}

The figures below illustrate grouped stiffness values (normalised to the average stiffness during self-selected hopping) for 5 participants performing sleigh hopping as per our instructions.

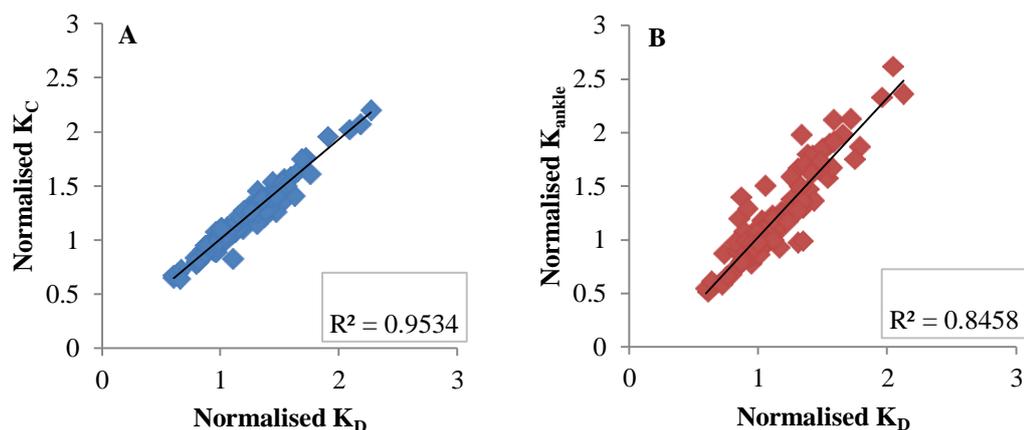


Figure A. The correlation between normalised lower limb stiffness calculated using the Cavagna (K_C) and Dalleau (K_D) methods for all five participants combined.

Figure B. The correlation between ankle joint stiffness (K_{ankle}) and normalised lower limb stiffness calculated using the Dalleau method (K_D) for all five participants combined.

With such high correlation values for both analyses we are satisfied that

1. K_D is valid for sleigh hopping, and
2. Under our specific methodology, K_D is measuring of ankle stiffness

Note: This investigation can be read in full as Supplemental Chapter 1 - Estimates of leg stiffness during low-load plyometrics (Grisbrook et al (Submitted Manuscript)).

3.4.3 Are we measuring the same domain?

The θ_{contact} (Pearson Correlation .610) and $\theta_{\text{take-off}}$ (Pearson Correlation .514) demonstrated a moderate correlation between SH and UH (Portney and Watkins 2009). Also, $\theta_{\text{c-p}}$ and stiffness demonstrated a strong to excellent correlation between UH and SH with a Pearson correlation of .817 and .887 respectively. These results suggest that stiffness, θ_{contact} , $\theta_{\text{take-off}}$ and $\theta_{\text{c-p}}$ are correlated across conditions. Clearly, stiffness was significantly higher ($p < .001$) for UH than for SH. However, the excellent correlation of stiffness and $\theta_{\text{c-p}}$ between conditions suggests that individuals have a default setting that they apply to each task i.e. a participants who produced a stiff performance during self-paced UH also produced a stiff performance when self-paced hopping on the sleigh. Thus, for these variables we are measuring the same domain for both upright and sleigh hopping.

Notably, θ_{peak} was not significantly correlated across conditions (Pearson Correlation = .364). The mean θ_{peak} for UH was 16 degrees dorsiflexion which represents a near end of range position and is likely is a function of the increased loading of upright hopping. Conversely, for SH ankle dorsiflexion had a mean θ_{peak} of 1.4 (+/- 5.8) degrees dorsiflexion recorded. It is plausible that for SH one does not need to move near end range to accommodate lower loading parameters. Therefore, it is unsurprising to the authors that this particular variable does not correlate across conditions.

3.4.4 Interpretation

Other researchers have employed various sleigh systems in order to collect SSC data in a controlled environment (Flanagan and Harrison 2007; Kramer, Ritzmann et al. 2010; Merritt, Raburn et al. 2012; Furlong and Harrison 2013). As outlined above, one cannot suggest that SH and UH are the same task. However, the highly reliable nature of lower limb stiffness and the ankle kinematic profile for SH and the moderate to strong correlations of the individual kinematic variables to UH justifies the use of SH as a means to model the repeated SSC of UH. Interestingly, when individuals self-regulate their hopping frequency they match their leg stiffness to the loading requirements of the task i.e. UH or SH. However, there exists a strong correlation between the stiffness and $\theta_{\text{c-p}}$ independent of the loading. Thus researchers may be able to collect large samples of sleigh hopping data and draw inference to normal dynamic function of the ankle. This may be particularly useful in research projects requiring large quantities of hopping data by providing a controlled, safe and low fatigue hopping environment. Furthermore, the standard error of measure presented

in this paper will allow for future studies with pathological groups and / or interventions and the assessment of meaningful change for each of our specific variables for both UH and SH.

3.5 Future research

Self-paced hopping was expressly utilised for both UH and SH conditions and unsurprisingly different kinematic profiles of the ankle were observed reflecting the loading requirements of each task. The loading requirements of SH could be constrained by adding external load or by predetermining the SH flight time and in doing so matching the work done to equal the work of UH. It is plausible that under such load-matched hopping that the kinematic profile of SH may more closely resemble the kinematic profile of SH.

3.6 Conclusions

The present paper compared leg stiffness and the kinematic profile of the ankle during self-paced upright and sleigh hopping. The results suggest that both are highly reliable tasks for examination of ankle stiffness and kinematic variables on a within and between day basis. The observed strong correlation between the stiffness and stretch amplitude independent of the loading suggests that individuals have a default behaviour that they apply to each task. Therefore, it is proposed that researchers may utilise self-paced sleigh hopping as an appropriate methodology for modelling normal dynamic function whilst negating the confounding features of upright hopping such as fatigue and balance.

4. The Minimal Perceptible Difference (MPD) Test

Travers M, Debenham J, Gibson W, Campbell A, & Allison G (2013) Stability of lower limb minimal perceptible difference in floor height during hopping stretch-shortening cycles. *Physiological Measurement*, 34(10), 1375 - 1386.

Stability of lower limb minimal perceptible difference in floor height during hopping stretch-shortening cycles

Mervyn J Travers^{1,3}, James Debenham^{1,2}, William Gibson^{1,2},
Amity Campbell¹ and Garry T Allison¹

¹ School of Physiotherapy, Curtin Health Innovation Research Institute, Curtin University, Kent Street, Bentley, WA 6102, Australia

² School of Physiotherapy, University of Notre Dame, 19 Mouat Street, Fremantle, WA 6959, Australia

E-mail: m.travers@curtin.edu.au

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Abstract

This study aimed to investigate a novel proprioceptive test, the minimal perceptible difference (MPD) test, that assessed participants' ability to perceive floor height changes whilst hopping. Sixteen healthy volunteers performed multiple trials of five hops on a custom built sleigh apparatus that permitted a floor height change (range 3–48 mm) or no change, as dictated by a structured searching algorithm. Minimum detected surface height change was recorded for eight different hopping conditions (factors–technique: alternate/bilateral hopping; side: dominant/non-dominant; direction of change: up/down) over two separate testing occasions. Within day and between day reliability were assessed using intraclass correlation coefficient (ICC) and 95% confidence intervals. The only factor which significantly influenced the sensitivity of subjects to detect changes in floor height was the hopping technique (bilateral or alternate). The mean MPD was significantly lower ($p < 0.0001$) for the bilateral hopping technique (MPD_{mean} = 15.7 mm) when compared to the alternate hopping technique (MPD_{mean} = 26.6 mm). All bilateral hopping techniques yielded moderate to high ICC for within (0.60–0.79) and between day (0.67–0.88) reliability. The results suggest that the bilateral hopping MPD assessment is a reliable, functional assessment of proprioception sensitivity during repeated stretch-shortening cycles that may better reflect human gait than established static assessment. Increased sensitivity to detection during bilateral hopping may reflect strategy dependent utility of proprioceptive information.

³ Author to whom any correspondence should be addressed.

Keywords: minimal perceptible difference, proprioception, sleigh hopping, stretch-shortening cycle

1. Background

The stretch-shortening cycle (SSC) of triceps surae acting across the ankle joint forms an integral basis of locomotion with specific neurological and physiological advantages over isolated muscle actions (Brughelli and Cronin 2008, Clark *et al* 2006, Fiolkowski *et al* 2005). This is particularly relevant to the operation of the foot and ankle where reflex contraction of the gastro-soleus complex, normal neurological drive, segmental energy transfer and the elastic recoil of the Achilles tendon are temporally integrated to contribute to every step in gait (Avela and Komi 1998, Horita *et al* 2003, Ishikawa and Komi 2007, Ishikawa *et al* 2006, Kallio *et al* 2004, Komi 2000, Ogiso *et al* 2002b). As such, an extensive body of research exists investigating the constituent components of the SSC (Ishikawa and Komi 2007, Ishikawa *et al* 2006, Komi 2000, Ogiso *et al* 2005, 2002b, Voigt *et al* 1998). The timing and synchronicity of the neurological component of the SSC is a dynamic process; (Avela and Komi 1998, Ishikawa and Komi 2007, Kallio *et al* 2004, Ogiso *et al* 2002b, Voigt *et al* 1998) for example the gastrocnemius stretch reflex has been demonstrated to be dependent on running speed (Ishikawa and Komi 2007). Also, the amplitude and latency of this stretch reflex varies depending on muscle length and level of activity (Kallio *et al* 2004, Ogiso *et al* 2005, 2002a, Voigt *et al* 1998) and is greatly affected by SSC fatigue (Avela and Komi 1998). The mechanical contributions of the passive and active components of the gastro-soleus complex to the SSC have also been examined, for example the contribution of elastic recoil appears to be dependent on the intensity of the SSC (Ishikawa *et al* 2006). Furthermore, the spring like function of the leg during repeated SSC has created interest in investigating stiffness—the resistance of a body or object to a change in length (Brughelli and Cronin 2008, Blickhan 1989). Like the stretch reflex, the contribution from active and passive components around the ankle to the ankle stiffness during the SSC is highly dependent on fatigue (Clark *et al* 2006, Komi 2000) and task demands (Chang *et al* 2008, Clark *et al* 2006, Ferris *et al* 1999, Grimmer and Blickhan 2006, Hobara *et al* 2007, Moritz and Farley 2005). Therefore, studies requiring repeated SSC have commonly utilized submaximal hopping as a model to investigate neuromuscular function (Voigt *et al* 1998, Dalleau *et al* 2004, Farley and Morgenroth 1999, Hobara *et al* 2007, 2008, Moritz and Farley 2004, 2005). Furthermore, sleigh systems have provided a reliable means of allowing investigation of repeated SSC whilst controlling for confounding factors such as fatigue and balance (Kramer *et al* 2010, Furlong and Harrison 2013).

To use the SSC in a safe and efficient way there has to be an optimal integration of feed-forward mechanisms (Bryant *et al* 2009, Moritz and Farley 2004, Taube *et al* 2012), sensory feed-back (Fiolkowski *et al* 2005), mechanical input (Komi 2000, Moritz and Farley 2005), landing mechanics (Morin *et al* 2007, Moritz and Farley 2005) and behavioural strategies (Leukel *et al* 2012). This complex interaction suggests that there is a strong internal integration of these different reflex, mechanical and conscious neurological drivers to modulate performance. Proprioception is a critical component of the sensory input required for movement (Proske *et al* 2000, Smith *et al* 2009, Winter *et al* 2005). To date the authors are unaware of any test which examines the proprioceptive component of the SSC. Therefore, a first step in creating a body of research in this area is to establish an appropriate methodology to investigate proprioceptive component of the SSC.

Proprioception is mediated by a complex synergy of sensory inputs which contribute to the formation of a person's postural schema and cerebral model of one's body size and shape

(Ivanenko *et al* 2011). As new input is perceived, it is compared to the existing map of body representation. In this way, the spatial localization of one's body may be continually updated and refined; these interacting processes are referred to as 'somatoperception' (Longo *et al* 2010). Furthermore, it has been posited that these sensorimotor inputs all converge in the spinal cord where the basic reflexive and rhythmic aspects of motor behaviour integrate with inputs from higher centres (Poppele and Bosco 2003). Therefore, it appears that proprioception is a much more complex concept than just joint position sense and kinaesthesia and this complexity may allow for significant redundancy in the proprioceptive system as no single input solely responsible for control or refinement of movement. Much of our understanding has been elicited by stimulating muscle spindle afferents via muscle and tendon vibrations and observing segmental and whole body changes in gait (Courtine *et al* 2007, Ivanenko *et al* 2000a) and both quiet and disturbed stance (Ivanenko *et al* 1997, 2000b). Importantly, this research model has identified a gating or utility of sensory input which may be dependent on the task being performed (Ivanenko *et al* 2000a). For example, application of vibration to leg muscles results in very different segmental and whole body changes between upright stance and treadmill walking (Ivanenko *et al* 2000a). This strategy dependent utilization of sensory input suggests that our ability to cognitively detect changes in our environment may also be task dependent as we may prioritize input which is critical to the individual's intended goal. This prioritization is worthy of consideration when devising a novel methodology for investigation of lower limb proprioception utilizing repeated SSC to model normal dynamic function. For this reason the minimal perceptible difference (MPD) test has been devised. The MPD test utilizes two different hopping styles (bilateral and alternate hopping) in a controlled environment (custom built sleigh) to allow the researchers to examine this strategy dependent proprioceptive redundancy across two different strategies that utilize repeated SSCs. The test entailed repeated trials of hopping performed on a low friction, custom built sleigh apparatus. The examiners measured the participants' ability to detect quantifiable changes in floor surface height during repeated SSCs.

This paper firstly aims to establish the reliability of the novel MPD test on a within and between day basis to serve as a foundation on which to further investigate this somatoperceptory performance. A secondary aim is to compare participants' ability to cognitively detect changes in floor surface height during repeated SSCs across different hopping strategies.

2. Methods

The MPD Test entailed repeated trials of hopping performed on a low friction, custom built sleigh apparatus (see figure 1). The sleigh was reclined to 20° and the weight of the sled (which participants reclined upon) was offset by a mass such that each participant was able to perform sub maximal hopping at a function of their own body weight. Furthermore, the sleigh reduced the impact of fatigue, balance and variance of centre of mass movement during hopping trials.

2.1. Participants

Sixteen healthy volunteers, seven female, nine male (mean age 25 years; range 18–35) were recruited for this investigation. Prior to participation in the study, participants signed an informed consent form that had been approved by Curtin University's Human Research Ethics Committee (approval number PT0145). Participants were excluded if they had any form of on-going pain or a history of lower extremity pathology, trauma, surgery or pain in the previous 2 years.

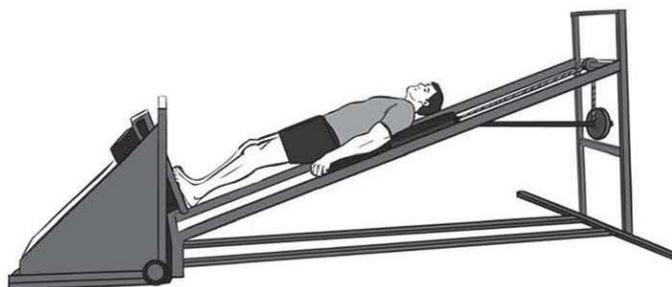


Figure 1. Custom built sleigh apparatus (Adapted from Gibson *et al* 2013).



Figure 2. The sliding floor mechanism illustrating the dominant bilateral down condition—this image shows a difference of 36 mm between the two feet.

2.2. Data collection protocol

Participants attended two data collections, one week apart, at Curtin University motion analysis laboratory. The data collection required participants to perform two different hopping techniques on the inclined sleigh—alternate hopping and bilateral hopping. Bilateral hopping required a simultaneous foot strike, while alternate hopping was characterized by alternating foot strikes. Following instruction on each hopping technique participants were provided a 10 min familiarization period to practice each hopping technique on the sleigh.

Following this, they were required to perform a series of trials of five hops. The participants had their eyes closed and were wearing noise cancelling headphones playing music to eliminate any visual or auditory feedback during testing. Therefore, to signal the commencement of a trial the tester would tap the participant on the leg. During each trial, one predetermined surface height change (upward or downward; minimum increment 3 mm, maximum increment 48 mm) was introduced under a predetermined foot (participant's dominant or non-dominant side). The custom built sleigh had an adjustable floor (sliding floor mechanism: figure 2) which allowed

for the rapid change of the floor height under either foot during the flight phase of a hop. A single surface height change was introduced in hops 2–5 in each block of five hops performed and the new floor height was left in place until the trial ended.

Participants were instructed to call out ‘change’ immediately when they perceived a change in surface height. A positive recognition of a surface change was registered if the subject shouted ‘change’ during the period from initial foot contact on a changed surface until the next foot contact of the hopping trial.

Importantly, comprehensive pilot testing revealed that participants were absolutely unable to detect surface changes accurately during sleigh hopping unless they were given a pre-instruction regarding the direction and side of change. This allowed participants to focus on the leg and direction of floor surface change for each trial. Furthermore, each participant was informed of two random assigned ‘ghost trials’ where no change was made were included for each hopping condition to attempt to control for guessing. Therefore, participants were given the following instructions to in order to manage the subjects’ expectation of a perturbation during a trial:

- Hop using predetermined technique (bilateral/alternate) at a comfortable pace.
- Tester may or may not make a floor change, any change would be under the foot and in the direction pre-determined e.g. non-dominant; up.
- Shout ‘change’ immediately if any perceived change in surface height.

2.3. Search algorithm

Pilot testing identified 18 mm as an appropriate starting surface height change for each condition. This initial 18 mm change was referred to as the ‘starting perturbation’. The magnitude of each subsequent floor height change was determined by our searching algorithms outlined below, and testing continued until the participant detected a change. There were two possible outcomes from the starting perturbation 18 mm—detected or undetected.

2.3.1. Detected starting perturbation. Following successful starting perturbation detection, the next trial employed a 9 mm increment. The subsequent trial either reduced or increased the increment by 6 mm depending on whether the 9 mm change was perceived or not e.g. 3 mm or 15 mm and followed the algorithm illustrated in figure 3 until a final score for that trial was obtained to a resolution of 3 mm.

2.3.2. Undetected starting perturbation. Following unsuccessful starting perturbation detection the next trial used a 36 mm change and, if detected the subsequent increments were reduced by 6 mm until the lowest detectable surface height change was detected using 6 mm increments i.e. 30 mm and then a 24 mm change. If the 36 mm change was not detected the subsequent increments were increased by 6 mm until the minimum detectable surface height change was measured i.e. 42 mm and then the maximum 48 mm change. The score for the trial was then obtained by testing with a final 3 mm increment removed. For example if, 36 mm was the lowest detected difference (in 6 mm increments) the participant was tested using a 33 mm change. Similarly, if 48 mm was the lowest detected difference (in 6 mm increments) the participant was tested using a 45 mm surface height change so that a final score for that trial was obtained to a resolution of 3 mm.

False statements during a no change trial were ignored with no feedback. Once three scores had been attained for each hopping condition they were averaged to give the participants final score the MPD in surface height. The test protocol took between 60 and 90 min including

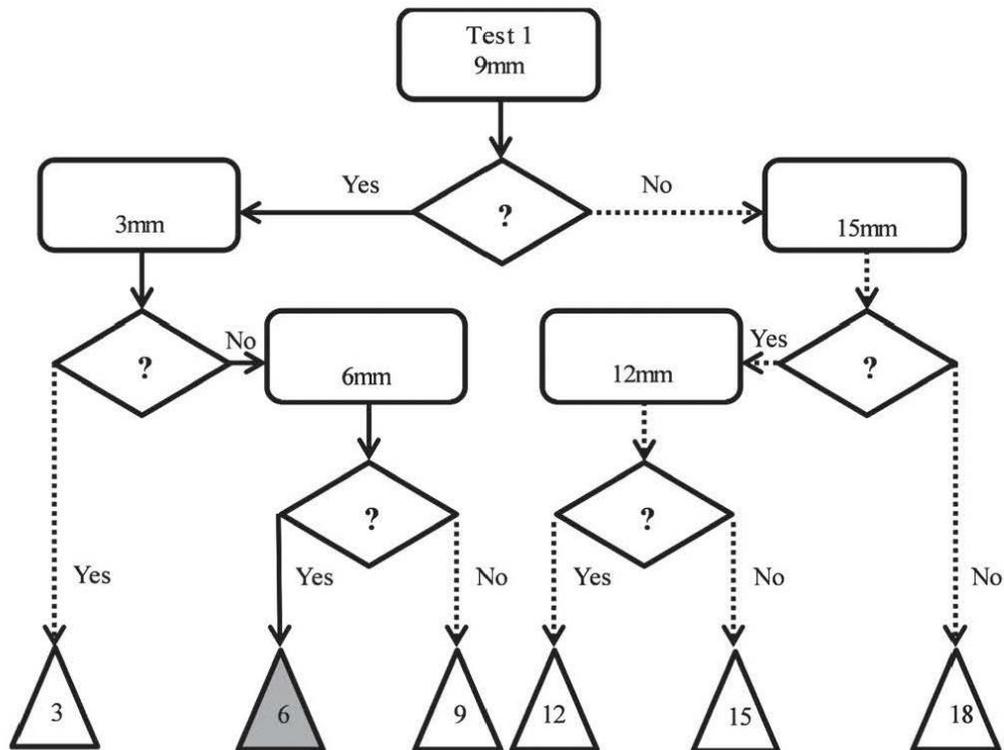


Figure 3. The detected starting perturbation search algorithm demonstrating an MPD score of 6 mm.

multiple rest periods—participants were given 10 s of rest between each trial and a 2 min rest after completing each condition. No participants reported any fatigue, discomfort or pain during testing.

2.3.3. Conditions. Each participant was tested using the above protocol under eight conditions that were generated by side (dominant/non-dominant), hopping technique (bilateral/alternate) and direction of change of surface height (up/down). These eight conditions were presented using a block balanced randomized order of testing. The sample size of 16 provided a balanced number of order of conditions within the study design. An *a priori* calculation of statistical power suggested that any detectable difference in main effects (dependent t-test post hoc comparisons) would be 95% confident in detecting a difference of 0.75SD of the population mean with a power of 80%.

3. Statistics

The MPDmean (in mm) of each condition was measured and input into statistical software for analysis (IBM SPSS Statistics version 2: IBM Corp©). Standard errors of measurement (SEM) with upper and lower 95% confidence intervals were used to represent the threshold for the MPDmean. Intra-class Correlation Coefficients (ICC) were derived to examine the reliability of the MPDmean for each condition, technique and day independently. A four-way repeated measures linear mixed model was used to identify any significant difference in

Table 1. Day 1 reliability of MPDmean scores during eight different hopping tasks.

Hopping condition			MPDmean (mm)	SD (mm)	SEM (mm)	95% CI range	ICC	95% CI range
Side	Technique	Direction						
Non-dominant	Alternate	Up	27.25	10.82	6.56	14.38–40.12	0.759*	0.541–0.899
		Down	27.38	10.15	6.81	14.02–40.73	0.601*	0.317–0.868
	Bilateral	Up	16.44	7.55	1.32	9.59–28.12	0.651*	0.383–0.864
		Down	16.50	4.86	3.63	9.39–23.61	0.685*	0.429–0.863
Dominant	Alternate	Up	23.07	8.89	1.37	12.48–42.63	0.570*	0.279–0.802
		Down	26.75	11.13	5.72	15.54–37.96	0.716*	0.474–0.878
	Bilateral	Up	14.32	6.61	1.24	9.42–21.78	0.774*	0.564–0.906
		Down	14.09	5.42	1.26	8.94–22.22	0.703*	0.445–0.872

SD = standard deviation; SEM = standard error of measurement; ICC = intraclass correlation coefficient;

* indicates $p < 0.0001$.

Table 2. Day 2 reliability of MPDmean scores during eight different hopping tasks.

Hopping condition			MPDmean (mm)	SD (mm)	SEM (mm)	95% CI range	ICC	95% CI range
Side	Technique	Direction						
Non-dominant	Alternate	Up	24.50	13.94	8.47	12.75–36.26	0.802*	0.610–0.918
		Down	23.94	10.12	4.91	14.31–33.57	0.740*	0.511–0.890
	Bilateral	Up	15.31	5.50	1.25	9.96–23.53	0.675*	0.416–0.858
		Down	13.69	7.57	3.17	7.48–19.90	0.792*	0.593–0.914
Dominant	Alternate	Up	27.41	11.51	1.34	15.52–48.41	0.605*	0.322–0.821
		Down	27.13	9.95	5.07	17.19–37.06	0.762*	0.545–0.900
	Bilateral	Up	13.52	5.57	1.31	7.915–23.10	0.600*	0.316–0.819
		Down	12.68	8.22	1.39	6.61–24.31	0.673*	0.413–0.857

SD = standard deviation; SEM = standard error of measurement; ICC = intraclass correlation coefficient;

* indicates $p < 0.0001$.

MPDmean grouped for condition, day and technique. Further, the linear mixed model was used to investigate any interaction between side, test day, direction of surface height change and hopping technique using the MPDmean as the dependent variable. An alpha of 0.05 was used to represent statistical significant for all comparisons.

4. Results

4.1. Within day reliability

The participants demonstrated moderate to strong reliability in MPD scores for bilateral hopping techniques on both test occasions, with slightly higher reliability when surface height changes were made on their dominant side (table 1). The dominant bilateral ‘up’ hopping task was the most reliable for assessing the MPD in surface height on a single test occasion. The mean MPD was 14.3 mm (95% CI 9.4–21.8 mm) on day 1. Furthermore, an ICC of 0.774 was yielded. The mean MPD was 13.5 mm (95% CI 7.9–23.1 mm) on day 2 and an ICC of 0.6 was yielded (table 2).

4.2. Between day reliability

Participants demonstrated moderate to strong reliability in MPD scores for bilateral hopping techniques between test occasions. The dominant bilateral ‘down’ hopping task was the most

Table 3. Between day reliability of MPDmean scores during eight different hopping tasks.

Hopping condition			MPDmean	SEM	95% CI range	ICC	95% CI range
Side	Technique	Direction	(mm)	(mm)			
Non-dominant	Alternate	Up	25.88	7.51	11.15–40.60	0.687*	0.104–0.891
		Down	25.66	5.00	15.85–35.46	0.765*	0.329–0.918
	Bilateral	Up	15.86	1.20	11.07–22.72	0.815*	0.472–0.935
		Down	15.09	3.92	7.40–22.78	0.669*	0.053–0.884
Dominant	Alternate	Up	25.14	1.16	18.67–33.86	0.898*	0.708–0.964
		Down	26.94	8.78	9.73–44.15	0.302	–0.997–0.756
	Bilateral	Up	13.92	1.23	8.47–88.86	0.668*	0.049–0.884
		Down	13.37	1.19	9.46–18.89	0.880*	0.656–0.958

SEM = standard error of measurement; ICC = intraclass correlation coefficient; * indicates $p < 0.05$.

Table 4. Fixed effects linear mixed model analysis using MPDmean as the dependent variable.

Source	Significance
Side	0.777
Technique	0.000*
Direction	0.437
Day	0.358
Side* technique	0.241
Side* direction	0.777
Side* day	0.114
Technique* direction	0.799
Technique* day	0.534
Direction* day	0.543
Side* technique* direction	0.562
Side* technique* day	0.344
Side* direction* day	0.777
Technique* direction* day	0.631
Side* strategy* direction* day	0.601

*Denotes a significant interaction with MPDmean scores.

reliable for assessing the MPD in surface height between test occasions. This task yielded a mean MPD score of 13.4 mm (95% CI 9.5–18.9 mm) and an ICC of 0.88 between test days (table 3). This between reliability was further verified as a linear mixed model analysis revealed no significant difference in MPDmean values between test days ($p = 0.431$).

4.3. Sensitivity

The results of the sensitivity analysis demonstrated that hopping technique (alternate/bilateral) was the only factor which significantly influenced the MPDmean values ($p < 0.05$) as demonstrated in table 4. The MPDmean value for the bilateral hopping technique was 15.7 mm (95% CI 14.1–17.2 mm). This was significantly lower than the MPDmean value for the alternate hopping technique which was 26.6 mm (95% CI 25.1–28.1). This demonstrated that subjects were significantly more sensitive at detecting surface height changes when hopping with simultaneous foot strikes. There was no significant effect on MPDmean values from side, direction of change or test day. Also, a four-way repeated measures fixed effects linear mixed model analysis using MPDmean as the dependent variable revealed no significant interaction between any of these factors ($p > 0.05$).

5. Discussion

This study has demonstrated the reliability and sensitivity of our proposed novel and functionally relevant tool; the MPD test. The current study represents a shift in focus from testing detection of passive positioning, position matching and force matching tasks of individual joints in proprioceptive focused research (Jong *et al* 2005, Lowrey *et al* 2010, Matre *et al* 2002).

5.1. Reliability

The ICC for bilateral hopping techniques ranged from 0.651 to 0.774 for day 1 and ranged from 0.600 to 0.792 on Day 2. Furthermore, the ICC for bilateral hopping ranged from 0.668 to 0.880 on a between day basis. These values exceed the proposed ICC of 0.6 that has been recommended for any measure to have clinical utility (Chinn 1991). Therefore, the results indicate that bilateral hopping techniques may have application in a research setting on single test and multiple test occasion experimental designs.

5.2. Sensitivity

Importantly, the only factor which significantly influenced the sensitivity of subjects to detect changes in floor height during the MPD test was the hopping technique employed. For this cohort, the MPD_{mean} was significantly lower for the bilateral hopping technique when compared to the alternate hopping technique. This suggests that participants were significantly ($p < 0.0001$) more sensitive at detecting the changes in floor height during bilateral hopping (MPD_{mean} = 15.7 mm) than during alternate hopping (MPD_{mean} = 26.6 mm). Our results demonstrate that sensitivity of detection of surface height changes during hopping is dependent on the hopping technique employed suggesting that we may possess different modes for movement detection for differing tasks. Thus, our results parallel previous findings which suggested that the gating of sensory input and utility of proprioceptive information differed between quiet stance and treadmill walking (Ivanenko *et al* 2000a). It may be hypothesized that the greater sensitivity we observed in the bilateral hopping tasks reflect the feedback and adaptation to bipedal standing tasks where very sensitive feedback for posture (standing) and initiation of movements is paramount. For stance, there may be no need for a proprioceptive memory to be held as there is constant simultaneous weight bearing feedback from both limbs i.e. a bilateral comparison contributing to one's postural schema. Whereas gait may require some kind of proprioceptive memory to be stored to cater for the lag time between alternating foot strikes.

The current knowledge base regarding lower limb proprioception and its influence on motor control has been based largely on tests of position matching and movement detection (Cameron *et al* 2008, 2009, Down *et al* 2007, Lowrey *et al* 2010). There is a consistent finding in the literature describing high degrees of sensitivity in these tests. For example, some tests of ankle inversion used in the experimental setting examine discrimination over a total range from 10 to 14 degrees inversion (Down *et al* 2007). Furthermore, proprioceptive ability can remain highly sensitive despite significant external challenges incorporating nociception and desensitization (Down *et al* 2007). This infers that humans may be highly sensitive at discriminating isolated movement at an individual joint and there is significant redundancy in the system allowing for sensitive detection of movement in spite of such challenges. However, it is plausible that testing position matching and movement detection does not represent the full complexity of proprioception and its influence on motor control (Ivanenko *et al* 2011, Longo

and Haggard 2012). It is unclear what factors determine the acuity of perception of changes in height of the floor surface during the bilateral SSC. Further investigation may determine if the MPD is related to a specific change in ankle angle at contact, total joint excursion, velocity of joint excursion, muscle force or leg stiffness.

Given the seemingly high precision of isolated joint movement detection it would be tempting to speculate this ability would be translated to similarly precise levels of discrimination in response to changes in floor height during hopping tasks. The mean floor height change detected during hopping by healthy subjects in our study was 26 mm for alternate hopping and 15 mm for bilateral hopping. Furthermore, we utilized cuing to prime participants in the direction and side of the potential surface height change during each trial. Without such cuing our piloted subjects were largely unable to detect surface height changes of 60 mm. Therefore, it may be hypothesized that humans exhibit a low degree of sensitivity to changes in the floor height when hopping. This may allow for large changes in the interface between the foot and the ground surface during gait without cognitive perception of the challenge. Thus, in a real world environment where surfaces are rarely homogenous we may be able to negotiate inconsistent surface heights during gait without disturbing efficient performance with too finely tuned perception of the incidental changes in limb length, leg stiffness or joint range that may be adopted to overcome these minor challenges. Therefore, until future research is able to document the association between dynamic and static ankle position sense testing then it is unclear if static tests are valid correlates to dynamic activities where sensory input may be gated depending on the task being performed.

5.3. Limitations

Whilst effort was made to minimize knee and hip flexion and to drive the movement from the ankles some movement naturally occurred throughout the leg and this could potentially introduce error. External bracing might appear to be a solution to unwanted movement; however this could have introduced another source of error through sensory input. The use of the sleigh allowed the researchers to reduce the impact of fatigue and balance on the SSC data collected, however it must be acknowledged that such controlled hopping performance is only a model for normal function and is not proposed to be the same task as gait or upright hopping. This study also utilized cueing to prime participants to the direction and side of the floor height changes, nonetheless large differences were still observed in detection of floor height changes between hopping strategies. Furthermore, participant variation may have been influenced by experience and expectations.

6. Conclusions

The MPD test has been presented as a novel test for assessing detection of floor height changes during sleigh hopping. This represents a change towards investigating the concept of proprioception in more functional tasks by using a repeated SSC model. Developing a robust and reliable method of quantifying this aspect of proprioceptive ability may allow for further investigation of functional proprioceptive ability in healthy and clinical populations. Furthermore, the MPD test has been demonstrated as reliable over time and is therefore an acceptable tool for assessing this function within and across test occasions. Significantly, greater sensitivity of the MPD test in the bilateral hopping technique may reflect finely tuned sensory requirements for upright stance which may be much less relevant for (bipedal) gait.

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5. Motor strategy modulation when expecting an external perturbation – a protective motor response or a sensory searching strategy?

5.1 Introduction

The critical interface during locomotion (walking, running, hopping) occurs between the foot and the ground surface. Adding to the complexity of this interaction is the fact that the ground surface is constantly changing representing a non-homogenous surface (van der Linden, Marigold et al. 2007; Grimmer, Ernst et al. 2008). Thus, negotiating varying external challenges is fundamental to the normal function of the lower limb. As such, understanding the motor control strategies governing lower limb performance in changing environments is important for clinicians and researchers engaged in human performance (Moritz and Farley 2005) and even robotics (Moritz, Greene et al. 2004). However, the foot is merely a part of a very complex system – the human neuromuscular system. In order to simplify gait, it has been often represented mathematically using a spring-mass model which describes the lower-limb behaving like a spring to progress the centre of mass (Blickhan 1989; Dalleau, Belli et al. 2004; Le Meur, Dorel et al. 2012). The stiffness of this leg spring refers to its resistance to deformation (Blickhan 1989; Dalleau, Belli et al. 2004; Le Meur, Dorel et al. 2012) and is provided by articular and periarticular tissues. We internally modulate this stiffness to specifically match a given task's demands (Yen, Auyang et al. 2009; Yeadon, King et al. 2010). In the research setting, hopping has commonly been utilised as a model for locomotion via its repeated stretch shortening cycles (SSC) (Farley, Blickhan et al. 1987; Blickhan 1989; Ferris and Farley 1997; Farley and Morgenroth 1999; Dalleau, Belli et al. 2004; Allison, Utsunoniya et al. 2005; Bobbert and Richard 2011) and has proven particularly useful for investigating stiffness modulation (Ferris and Farley 1997; Moritz and Farley 2004; Moritz, Greene et al. 2004; Moritz and Farley 2005; Arya, Solnik et al. 2006; Hobara, Omuro et al. 2007; Chang, Roiz et al. 2008; van der Krogt, de Graaf et al. 2009; Bobbert and Richard 2011). Importantly, it is demonstrated that the ankle is largely responsible for stiffness regulation during submaximal SSC activity (Moritz and Farley 2004) and the SSC component of this activity can be largely isolated using a sleigh hopping model (Furlong and Harrison 2013; Travers, Debenham et al. 2013) (Grisbrook et al, Submitted Manuscript – see Supplementary Chapter 1; Debenham et al, Submitted Manuscript- see Appendix 1) and as such is utilised in this current study.

The existing literature suggests that humans are well equipped to deal with both internal (Unpublished Data – see Supplementary Chapter 2) and external challenges to leg stiffness

during submaximal hopping. For example, participants performing a hopping task have been shown to modify their mechanics and muscle activity profiles in order to maintain centre of mass dynamics when hopping on a range of surface stiffnesses (Moritz, Greene et al. 2004) and specifically on highly compliant and elastic surfaces (Moritz and Farley 2005). For example, when hoppers anticipated a surface stiffness change they adjusted both their limb kinematics and increased leg muscle pre-activation in order to prepare for the expected change in stiffness demands (Moritz and Farley 2004). A key feature of this body of knowledge is the focus on the protective feed-forward responses (Taube, Leukel et al. 2012) used to modify leg stiffness to cater for expected external challenges during hopping where there is risk of sub-optimal performance, falling or injury associated with control decisions – i.e. perturbed performance (Moritz, Greene et al. 2004), reduced sensory input (Fiolkowski, Bishop et al. 2005; Fu and Hui-Chan 2007) or expectation of pain (Moseley, Nicholas et al. 2004). However, it is currently unknown what role expectation of a changing environment may play in participants' baseline hopping strategy.

Forward models of motor control suggest that humans continually produce internal models of predicted movement and compare to actual performance (Miall and Wolpert 1996; Blakemore, Goodbody et al. 1998). This forward model has been thoroughly investigated in gait where short latency neural responses to unexpected changes in floor height were observed (van der Linden, Marigold et al. 2007). Importantly, when these participants expected a change in floor height they walked with a more plantarflexed ankle at heel strike even if no change occurred, suggesting a motor priming based on expectation of a change in the environment (van der Linden, Marigold et al. 2007). Thus, when examining stiffness modulation in response to external challenges using a hopping model, it is plausible that participants may change their motor strategies in response to the knowledge alone that they are operating in an unpredictable environment. There exist no other studies which purely describe the role of expectation of a changing external environment on the baseline mechanics and neural drive in hopping performance.

This paper aims to answer the following research questions:

1. Does expectation of a change in surface height during bilateral hopping alter the sagittal kinematic profile of the ankle?
2. Does expectation of a change in surface height during bilateral hopping alter the EMG profile of the ankle?

5.2 Methods

5.2.1 Participants

This study utilised a within-subject experimental design. Following informed consent (HREC approval #PT0189) participants attended one testing session at the Motion Analysis Laboratory, Curtin University. Eight healthy participants (3 male, 5 female; mean (SD) age 31 (2), height 172.4 (9.1) cm, body mass 72.8 (15.7) kg) free of any pain or functional limitations were tested. All participants were right foot dominant.

5.2.2 The Sleigh - Providing a risk minimised environment

Participants performed multiple trials of double legged hopping on a custom-built sleigh apparatus (Figure 5.1) (Gibson, Campbell et al. 2013). This low friction sleigh system was reclined to 20 degrees from horizontal to create a controlled, low fatigue hopping environment (Debenham et al, Submitted Manuscript- see Appendix 1) with the weight of the sliding board offset by an external mass and pulley system. It is a useful tool for examining stiffness modulation (Unpublished Data – see Supplementary Chapter 2) and allows for focus on performance at the ankle (Grisbrook et al, Submitted Manuscript – see Supplementary Chapter 1). Critically, the use of the sleigh was intended to negate balance / risk of falling as a feature of maintaining centre of mass dynamics in a low load environment.

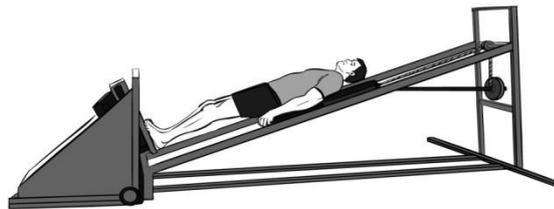


Figure 5.1 Double leg hopping on the custom built sleigh apparatus set at 20 degrees inclination with a mass offset of the low friction sled

Image Reproduced from (Gibson, Campbell et al. 2013).

5.2.3 Procedure

The participants were requested to perform multiple hopping trials under two conditions - Baseline Hopping (BH) and Expectation Hopping (EH). BH entailed three trials of 10 continuous bilateral hops at the participants' self-selected frequency and natural ground contact time.

For the EH trials, a sliding floor mechanism (Figure 5.2) was attached to the base of the sleigh. This allowed the researchers to adjust the height of the landing surface during the flight phase of a hop. The participants performed six trials of 5 hops at their preferred frequency and ground contact time. One predetermined surface height change was made in the upwards direction under the right foot (all participants were right foot dominant) on either the second, third or fourth hop during each EH trial. The randomised floor height changes were of either 6mm or 36mm as previous research suggested that participants would likely not perceive or would perceive them respectively (Travers, Debenham et al. 2013). Participants were instructed in advance that an upwards change in surface height would happen during each EH trial, but they may or may not perceive the changes and were instructed to shout 'change' to notify the testers perception. Hopping was performed whilst minimising knee flexion (no external fixation was used), and participants kept their eyes closed and were wearing headphones with background music to eliminate auditory feedback.

In a previous study, the threshold for detection of changes in surface height during sleigh hopping has been investigated (Travers, Debenham et al. 2013). The Dominant Bilateral Up condition, meaning that any changes were made in the upward direction, under the dominant foot whilst bilaterally hopping was utilised for the present study as it was demonstrated to have high reliability on a within day basis ($ICC = 0.774$) (Travers, Debenham et al. 2013).

5.2.4 Familiarisation and Controlling for Risk-Driven Expectation

A ten minute familiarisation period was conducted for all participants to become comfortable with both hopping conditions and the use of the sleigh. In particular, familiarisation for the EH condition was performed with both eyes closed (replicating the test proper) and with eyes open so that participants could be sure that floor height changes were real and of little risk, irrespective of perception.



Figure 5.2 The Sliding Floor Mechanism

This image shows a difference in landing surface height of 36mm between the two feet (Dominant Bilateral Up)

5.2.5 3-D Motion Analysis

Fourteen Vicon (Oxford metrics, Inc.) infra-red cameras, sampling at 250Hz, captured the movement of reflective markers that were applied to the lower limbs as per Chapter 3 – Upright vs Sleigh Hopping.

5.2.6 Identifying Event Markers during hopping using 3-D Motion Analysis

Traditionally, force plates represent the gold standard for identifying event markers in gait analysis (Cavanagh and LaFortune 1980; Buczek, Cavanagh et al. 1991; O' Riley, Dicharry et al. 2008; Lieberman, Venkadesan et al. 2010). However for this study a sliding floor mechanism (Figure 5.2) was attached to the force plate of the sleigh allowing adjustment of the landing surface height during the flight phase of a hop. Although this represented an ideal method for adjusting the floor height during a hopping trial, the interface between the force plate, adjustable wooden rig and participants' foot rendered accurate measurement of hopping contact and flight phases with the force plate impossible. Therefore, standard event markers such as foot contact and toe-off for both left and right feet independently during bilateral hopping were established using the Vicon 3-D Motion Analysis System. The efficacy/validity of identifying these events using the 3-D motion analysis has been compared with the known gold standard (force plate) and demonstrated to be very reliable, yet significantly different (Unpublished data – See Supplementary Chapter 2). This

difference was systematic and demonstrated to have no effect on stiffness estimates as confirmed by both a very high reliability (ICC = 0.96) between stiffness estimated with the two different event identification methods (Unpublished data – See Supplementary Chapter 2). The systematic difference equates to a 4.3% overestimation of stiffness when using the Vicon 3-D Motion Analysis System to establish the phases of hopping (Unpublished data – See Supplementary Chapter 2).

5.2.7 Electromyography (EMG)

An AMT-8 (Bortec Biomedical Ltd.) system was utilised to collect surface EMG signals bilaterally from the medial gastrocnemius (MG), soleus (SOL), and the tibialis anterior (TibAnt) muscles. Bipolar differential surface electrodes (Ag / AgCL) were placed on the belly of each muscle with the reference electrode on the tibial plateau. Skin impedance (< 15 kOhms) was achieved with skin preparation and signals were pre-amplified, analogue filtered (10 – 500Hz band pass) and then digitised using an 18 bit A-D card with a sampling rate of 1000Hz. All data was temporally synchronised and recorded to hard disc on the Motion Laboratory dedicated hardware running a customised Labview program (National Instruments Inc.).

5.2.8 EMG signal Onsets, Normalisation and Conditioning

The EMG data were full wave rectified and onsets detected using the integrated protocol (Allison 2003). Trial linear envelopes (LE) were created using a fourth-order, zero-lag Butterworth low-pass filter (10 Hz) and temporally synchronised to (T=0) foot contact. Ensemble average LE were determined for a 760 ms window defined as 280 ms prior to contact and 480 ms after contact. This is consistent to detect medial gastrocnemius pre-activation onsets (Jones and Watt 1971). EMG signals were integrated in 20 ms epochs (IEMG) for the 760 ms window. EMG data was also time normalised to the duration of the contact phase for secondary analysis. Amplitude normalisation was undertaken with the median peak of the submaximal (Allison, Marshall et al. 1993) hopping familiarisation trials (10 consecutive hops) used as 1.0 arbitrary unit (A.U).

5.2.9 Leg Stiffness

Leg Stiffness was estimated using the method proposed by Dalleau et al (2004) as described in Chapter 3 – Sleigh vs Upright.

5.2.10 Kinematic Variables

As per Chapter 3 – Sleigh vs Upright, the following kinematic variables were outputted for each leg during the bilateral task:

1. Ankle angle at contact in sagittal plane (θ_{contact})
2. Peak dorsiflexion angle (θ_{peak})
3. Stretch Amplitude ($\theta_{\text{c-p}}$): the change in the ankle joint angle from landing (contact) to the most dorsiflexed point
4. Ankle angle at take-off θ in sagittal plane ($\theta_{\text{take-off}}$)

Additionally, the stretch velocity (θ_{vel}) in degrees/ second ($^{\circ}/\text{s}$) was defined as the mean dorsiflexion angular velocity, equating to the ratio of the dorsiflexion amplitude to the time interval from landing to the most dorsiflexed point.

5.2.11 Analysis

We analysed the data from two hops preceding the change in floor height from each of six trials, resulting in 12 hops for the EH condition per participant. For the BH condition we analysed the data from the equivalent two hops for each of three trials giving 6 BH hops per participant. Kinematic data was analysed for eight participants, with EMG data from six participants due technical error and subsequent signal loss. All Data analysis was performed using statistical software (IBM SPSS Statistics version 20: IBM Corp©). An alpha of 0.05 was used to represent statistical significance for all comparisons and all comparison were set apriori. The null hypothesis that prior knowledge of an impending perturbation would not affect motor performance during hopping was tested using a series of paired samples t-tests to determine significant differences in mean values across the two conditions for all variables. As the participants were performing a bilateral hopping task, data for the left and right feet are presented.

5.3 Results

5.3.1 Kinematic Data

The participants demonstrated a statistically significant decrease in contact duration for both right and left legs during EH in comparison to BH ($p < 0.020$) (See Table 5.1). There was a concurrent increase in stiffness of the right leg for EH ($p < 0.014$). Table 5.2 demonstrates the sagittal plane ankle angle (in degrees) at specific time points during the contact phase for BH and EH. Table 5.3 demonstrates the differences in θ_{c-p} and θ_{vel} across conditions. Figure 5.3 illustrates the time normalised sagittal plane ankle profiles during the BH and EH conditions for both ankles.

Table 5.1 Comparison of the descriptive variables between BH and EH conditions

Variable	Side	BH mean (SD)	EH mean (SD)	Mean Diff	95% CI	p
Hopping Frequency	R	1.30 (.15)	1.32 (.09)	-0.02	-.15 to .11	.734
	L	1.32 (.10)	1.30 (.14)	.02	-.09 to .13	.705
Contact Duration (ms)	R	396 (32)	356 (30)*	39	12 to 68	.020
	L	390 (32)	359 (25)*	31	6 to 54	.018
Flight Duration (ms)	R	426 (149)	427 (67)	0.4	-144 to 144	.995
	L	408 (92)	434 (109)	-26	-136 to 83	.558
Stiffness (kN.m-1)	R	8.80 (2.65)	10.21 (3.07)*	-1.4	-2.42 to -.39	.013
	L	9.71 (3.93)	9.92 (2.66)	-.21	-1.42 to 1.01	.701

*Denotes a significant increase between BH and EH conditions; Mean Diff = Difference in Means; 95% CI = 95% Confidence Interval of the difference

Table 5.2 Comparison of the sagittal plane ankle angle (in deg) at specific time points during the contact phase for BH and EH conditions for both legs

Variable	Side	BH mean (SD)	EH mean (SD)	Mean Diff	95% CI	p
θ_{contact}	R	-21.8 (5.5)	-23.0 (5.4)	1.2	-3.5 to 5.8	.572
	L	-19.2 (5.2)	-24.1 (4.3)*	4.9	1.1 to 8.8	.019
θ_{peak}	R	3.9 (6.2)	4.7 (4.6)	-0.8	-6.5 to 4.9	.755
	L	5.2 (4.7)	3.8 (5.2)	1.4	-2.8 to 5.6	.456
θ_{take-off}	R	-31.1 (8.2)	-34.4 (6.6)*	3.3	0.3 to 6.3	.035
	L	-30.2 (6.7)	-35.4 (5.6)	5.2	-1.5 to 11.8	.109

*Denotes a significant increase between BH and EH conditions; Mean Diff = Difference in Means; 95% CI = 95% Confidence Interval of the difference; Minus values denote plantarflexion

Table 5.3. Comparison of stretch amplitude and stretch velocity for BH and EH conditions

Variable	Side	BH mean (SD)	EH mean (SD)	Mean Diff	95% CI	p
θ_{c-p} (°)	R	25.7 (6.9)	27.7 (4.5)	2.0	1.33 to 5.23	.203
	L	24.5 (5.9)	27.9 (4.1)*	3.4	.13 to 6.9	.044
θ_{vel} (°/s)	R	264.1 (71.0)	344.2 (48.3)*	80.1	43.2 to 110.9	.001
	L	276.1 (79.3)	337.6 (57.1)*	61.5	12.8 to 110.1	.020

*Denotes a significant increase between BH and EH conditions; Mean Diff = Difference in Means; 95% CI = 95% Confidence Interval of the difference

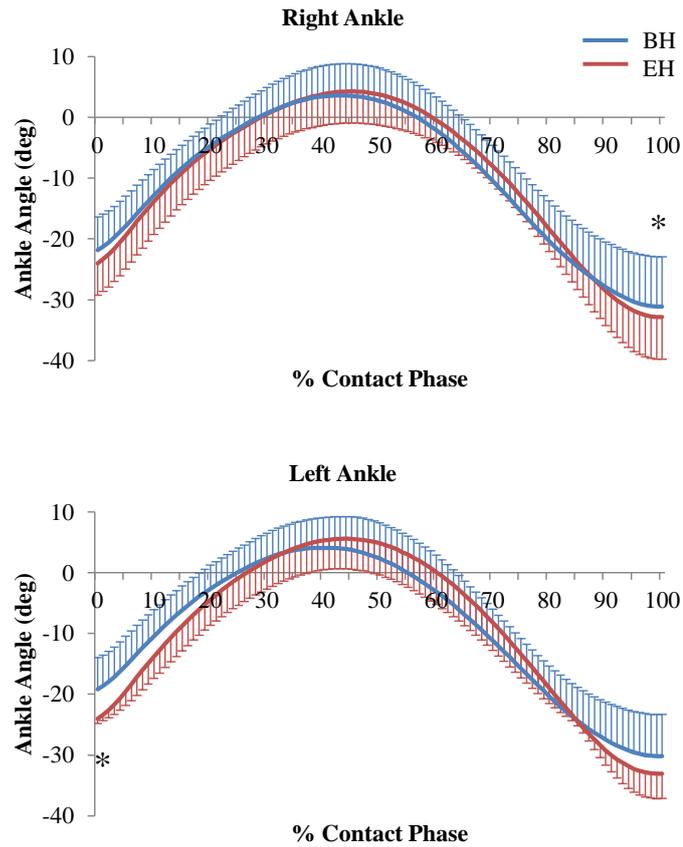


Figure 5.3. Time normalised, group mean sagittal plane ankle profiles with standard error (error bars) during the BH and the EH conditions

Both graphs have been normalised to contact phase duration where 0 = contact, 100 = toe-off / end of the contact phase (X-Axis); Y-Axis: ankle angle; positive values = Dorsiflexion; Minus values = Plantarflexion; * indicate ranges where a significant difference existed between conditions; Error bars represent standard error (n=6)

5.3.2 EMG Data

Table 5.2 outlines the EMG onsets and time to peak (in ms relative to contact) for BH and EH. No significant differences were observed for any of these discrete EMG variables between conditions (all $p > .293$).

Table 5.2 EMG onsets and time to peak (in ms relative to contact) for BH and EH

	Right			Left		
	BH	EH	p value	BH	EH	p value
	Mean (SD) 95% CI	Mean (SD) 95% CI		Mean (SD) 95% CI	Mean (SD) 95% CI	
Soleus						
Onset	79 (24) 61 to 99	77 (33) 50 to 103	.775	78 (39) 47 to 109	66 (40) 34 to 98	.334
Time to Peak	198 (25) 177 to 218	114 (19) 101 to 125		.723	189 (36) 161 to 217	
Medial Gastrocnemius						
Onset	37(19) 22 to 52	20 (45) -15 to 57	.404	44 (20) 18 to 69	24 (33) -2 to 50	.364
Time to Peak	204 (25) 184 to 224	176 (39) 145 to 208	.293	181 (36) 152 to 209	184(39) 153 to 215	.885
Tibialis Anterior						
Onset	63 (80) -1 to 137	36 (31) 9 to 59	.338	68 (92) -5 to 141	42 (22) 24 to 59	.532
Time to Peak	177 (52) 135 to 218	175 (48) 137 to 213	.953	188 (84) 120 to 255	183 (36) 153 to 211	.904

95% CI = 95% Confidence Interval of the mean; Negative values refer to duration prior contact. BH = Baseline Hopping, EH = Expected Hopping

Figure 5.4 illustrates the time normalised EMG profiles for the Sol, TibAnt and MG during the contact phase for both conditions and ankles. Each muscle had its own specific window where the EMG activity significantly increased for the EH condition. Importantly, there was a specific window from .25 to .425 of the contact phase where all six channels had a significant increase for the EH condition for each data point in that range (all $p < .05$).

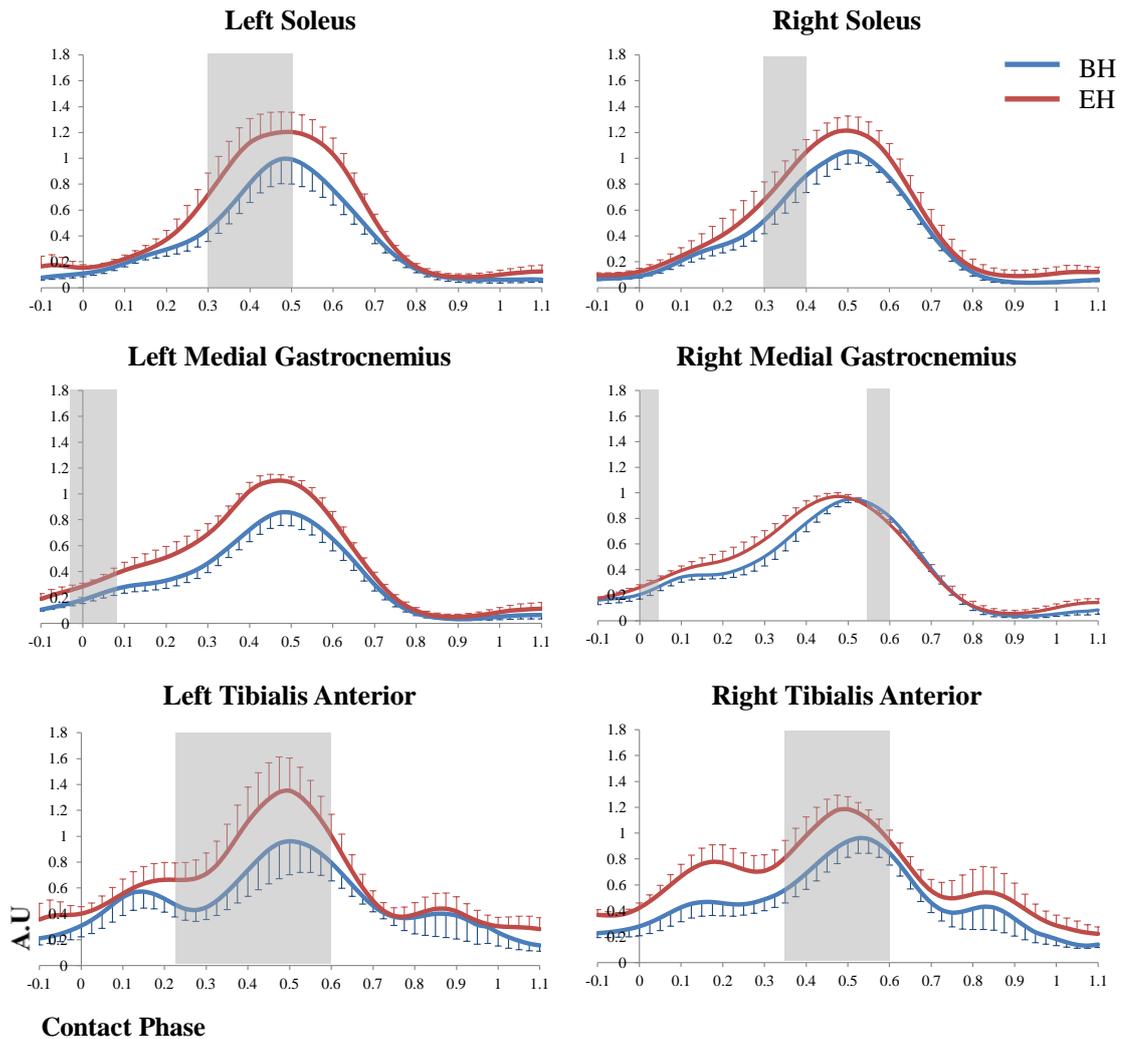


Figure 5.4. Time normalised mean linear envelopes for the Sol, TibAnt and MG muscles during the BH and EH conditions

All graphs have been normalised relative to the contact phase duration where 0 = contact, 1 = toe-off / end of the contact phase and minus values indicate pre-contact; X-axis: time (s); Y-Axis: the 1.0 arbitrary unit (A.U) reflects the peak value for the familiarization protocol; Shaded areas represent ranges where a significant difference existed between conditions; Error bars represent standard error (n=6).

5.4 Discussion

Current research explains the mechanical and neural adjustments used to modulate leg stiffness in response to both expected and unexpected external challenges during hopping (Moritz and Farley 2004; van der Krogt, de Graaf et al. 2009). However, these observations have been made by comparing the participants' baseline hopping strategy to their performance on the new surface. Such an experimental paradigm does not account for the

role that expectation alone of a changing environment may play in altering participants' baseline hopping strategy. This study has attempted to quantify this effect.

We observed a systematic decrease in contact duration for both left and right legs ($p < 0.020$) with a concurrent increase in stiffness of the right leg ($p = 0.013$) when participants expected an increase in landing surface height under the right foot (Table 5.1). It has been reported previously that when participants double-leg hopped on a soft surface which became expectedly stiffer they reduced their contact time and increased leg stiffness significantly when landing on the expected stiff surface (Moritz and Farley 2004). We have demonstrated this change in contact time may be attributable to the expectation of the surface change alone. We observed the increase in leg stiffness for the right side only; it is plausible that this unilateral modulation may have been dictated by priming the participants to the side of the change (right side). It may be that altering stiffness during EH is integral to detecting the impending change.

The kinematic profile of the ankles changed very little when compared across conditions (Figure 5.3). The right ankle was significantly less plantarflexed ($p < 0.05$) for the final 2.5% of the contact phase during EH. Conversely the left ankle was significantly more plantarflexed ($p < 0.05$) during the first 2.5% of the contact phase. The stretch amplitude for the left leg was increased by 3.4 degrees for EH representing a statistically significant difference ($p = 0.044$) (Table 5.3). No other differences in sagittal ankle profile were observed across conditions for either side; however a significant increase in θ_{vel} ($p < 0.020$) was observed bilaterally for EH (Table 5.3). It is unclear whether this increase in stretch velocity for EH represents a strategy to increase leg stiffness or whether it is an attempt to optimise proprioceptive sensitivity via the stretch-velocity relationship (Ribot-Ciscar and Roll 1998; Proske, Wise et al. 2000). Further research is required to explain the relationship between stretch velocity and proprioceptive acuity during the SSC.

As illustrated in Figure 5.4, expectation produced statistically significant increases in EMG activity for the Sol, MG and TibAnt muscles for both legs (all $p < .05$). Notably, for each individual muscle there appears to be a common window where this commences for both sides. For example the window of significant increase in Sol muscle activity commences at 30% of the contact phase bilaterally. A similar observation is true for the mono-articular antagonist TibAnt muscle where this window commences at 25% and 35% of the contact phase for left and right legs respectively, with the window ending at 60% of the contact phase for both legs. The timing and duration of the above increases in EMG activity for the

EH condition appears to be consistent with the period of peak dorsiflexion occurring between 40 and 45% of the duration of contact as outlined in Figure 5.3. It is important to note that these changes in muscle activation occurred despite there being no significant difference in flight time, and thus work done for either leg during EH when compared to BH (Right $p = .995$; Left $p = .558$). Furthermore, these increases in muscle activation are unlikely due to an enhanced stretch reflex as the change was observed in both agonist and antagonist channels (Berardelli, Hallett et al. 1982). It is also an interesting observation that these changes occurred under both feet when the participants were only primed to a change under the right foot. Further research making changes under left and right feet independently may indicate whether this is a bilateral expectation response irrespective of the side of the impending perturbation. Future research should consider if changing the side of the expected perturbation changes the adaptive response. The change in stiffness (unilateral) may support this inference, however the other domains that were assessed in the present study do not support this hypothesis.

A different pattern has been observed for the MG muscle where bilateral increases in muscle activity occurred close to the point of contact for both sides (2.5% pre-contact for the left leg and at the contact point for the right leg). This is more consistent with previous findings of increased activation of multiple leg muscles including Sol, MG and TibAnt prior to contact when hoppers expected an increase in floor stiffness (Moritz and Farley 2004). A key distinction however is that our participants were operating in a low load, risk minimised environment where the expected perturbations were familiar, safe and subtle. The intention was not to maintain centre of mass dynamics in the face of an external perturbation. We utilised a different research paradigm where the primary instruction to the participants was to search for the subtle perturbation that they may or not feel. In fact, we primed them to the direction (up or down) and the side of the surface change (right). It cannot be discounted that the different EMG profiles we observed for the expectation condition between the bi-articular MG and the mono-articular Sol and TibAnt may be due to the participants' efforts to try to perceive the impending perturbation. Therefore, the observed changes in EMG signal amplitude described above may actually represent a sensory searching strategy rather than the traditionally described motor strategies that rationalise the adaptive responses to a mechanical optimization approach. The findings of this paper suggest that there could be both sensory and mechanical optimization strategies that are modulated according to the level of risk perceived or the magnitude of the pending perturbation.

In the present study, the motor behaviour of the ankles when expecting a familiar, safe and subtle perturbation was examined. Furthermore, the perturbation occurred during SSC activity in a low load / risk environment using the custom built sleigh apparatus where the participants' body weight was offset and the potential for loss of balance or failed loading was minimised. With similar studies in walking a potential 'foot in hole scenario' introduces the real risk of a fall / largely disturbed performance. In two previous studies utilising expected and unexpected variations in floor height during walking, the participants walked with increased plantarflexion at heel contact when they thought a change in floor surface may occur (van der Linden, Marigold et al. 2007; Masahiro and Shingo 2010). This is consistent with our observation of changes in the kinematics of bilateral hopping due to an expectation of change in floor height. Even greater risk of failed loading exists in studies which have examined the feedforward control of drop landings and drop jumps (McDonagh and Duncan 2002; Leukel, Taube et al. 2012). These studies have demonstrated the feedforward control of SSC activity and the associated modulations to cater for an expected perturbation it is impossible to separate the risk of failed loading (injury or a fall) when performing drop jumps or landings (McDonagh and Duncan 2002; Leukel, Taube et al. 2012). The present study represents the first time that this phenomenon has been investigated under such controlled circumstances and demonstrates that feedforward modulation for the SSC exists even under low-load / risk conditions.

It has been suggested that when participants expect a floor height change during walking trials, they may predict the change in height would be detectable specifically at the point of heel contact and that the cerebellum might prime spinal structures to detect the perturbation at this critical point in gait (van der Linden, Marigold et al. 2007). In the present study, the participants were primed to the side of change (right), direction of height change (upwards) and were searching for the change in surface height at effectively the same point during the hopping cycle (toe contact). Therefore, it may be hypothesised that they may have been searching for the change in floor height at toe contact. Thus, a priming of spinal structures via the cerebellum may have driven the systematic EMG signal increases observed across all channels and on both legs during the contact phase of EH. Future research should introduce unpredictable changes to the floor height throughout the contact phase to see if similar observations to the present findings are produced. To specifically examine the role of the cerebellum in such anticipatory strategies could be investigated by testing patients suffering from cerebellar dysfunction.

5.5 Conclusion

In conclusion, expectation of a change in the foot / landing surface interface produced a change in baseline bilateral hopping strategy. These changes included reduced contact time and increased stretch velocity bilaterally, and an increase in stiffness in the leg on the side of the expected change. EMG signal amplitude increased in both agonist and antagonist muscles of both ankles in response to expectation of a change in surface height under a single foot. Different activation patterns observed for the mono-articular and bi-articular muscles may represent a sensory searching strategy rather than a motor response to cater for the impending perturbation. Future studies utilising external perturbations during repeated stretch shortening cycles should account for the role of expectation alone on baseline hopping strategies.

6. Does perception of an expected perturbation induce a motor strategy response?

6.1 Introduction

The normal function of the lower limb is to progress the centre of mass safely and with optimal efficiency. The repeated stretch shortening cycle (SSC) seen around the ankle joint in normal locomotion delivers an enhanced efficiency beyond the outputs of isolated muscle actions (Fiolkowski, Bishop et al. 2005; Clark, Millard et al. 2006; Brughelli and Cronin 2008). In this way the ankle joint acts as a key contributor to every step via optimised integration of reflex contraction of the gastro-soleus complex, normal neurological drive, segmental energy transfer and the elastic recoil of the Achilles tendon (Avela and Komi 1998; Komi 2000; Ogiso, McBride et al. 2002; Horita, Komi et al. 2003; Kallio, Linnamo et al. 2004; Ishikawa, Komi et al. 2006; Ishikawa and Komi 2007). Optimisation of the effect of the SSC requires integration of feed-forward mechanisms (Bryant, Newton et al. 2009; Taube, Leukel et al. 2012), mechanical input (Komi 2000), landing mechanics (Moritz and Farley 2005; Morin, Samozino et al. 2007), behavioural strategies (Leukel, Taube et al. 2012) and sensory feed-back (Fiolkowski, Bishop et al. 2005).

Proprioception is a critical component of the sensory input required for this complex internal integration and execution of optimal movement (Smith, Crawford et al. 2009; Proske and Gandevia 2012). In the case of active movement, research on the specialised receptors contributing to proprioception has focused on the muscle spindle in particular (McCloskey 1973; Gandevia and McCloskey 1976; McCloskey, Cross et al. 1983; Gandevia 1985; Colebatch and McCloskey 1987; Proske, Morgan et al. 1993; Wise, Gregory et al. 1996; Ribot-Ciscar and Roll 1998; Wise, Gregory et al. 1998; Winter, Allen et al. 2005; Windhorst 2007). However, in order for proprioceptive input from spindles or other receptors to actually reach conscious perception there needs to be a discrepancy between the predicted sensory feedback and the actual sensory feedback (Bays and Wolpert 2007; Proske and Gandevia 2012). Recently, the Minimal Perceptible Difference (MPD) test has emerged as a proprioceptive test which specifically examines participants' ability to detect changes in floor surface height during repeated hopping trials (Travers, Debenham et al. 2013). The existing literature suggests that humans can adequately cater for such changes in their environment via stiffness modulation (Ferris and Farley 1997; Ferris, Liang et al. 1999; Moritz and Farley 2004; Müller and Blickhan 2010). For example, participants have been shown to adjust their mechanics and muscle activity profiles in order to maintain centre of mass dynamics when hopping on a range of surface stiffness (Moritz, Greene et al. 2004).

Another cohort adjusted their limb kinematics and increased leg muscle pre-activation when anticipating a surface stiffness change when hopping (Moritz and Farley 2004). Central to such research paradigms is the perturbation of performance and examination of the associated biomechanical and motor responses to the challenge / change in environment (Ferris and Farley 1997; Moritz and Farley 2004; Moritz, Greene et al. 2004; Moritz and Farley 2005). Importantly, these studies do not consider the influence of cognitive perception of a change in the environment on participants' hopping performance. However, it appears from the MPD test that there is a specific threshold at which participants become aware of a change in floor height during a particular hop cycle (Travers, Debenham et al. 2013). Theoretically, a 'sensory discrepancy' must occur at this threshold perturbation in order to trigger conscious perception (Bays and Wolpert 2007). This raises the question of whether the same performance changes occur for a subliminal perturbation as for one that is above the threshold of perception. Therefore, it is unknown whether motor strategy responses to changes in the environment are a direct result of the introduced mechanical change or does one's cognitive perception of the change influence the behaviour. To this end the authors have utilised the MPD test to answer the following question:

- When perturbations of a magnitude that are on either side of the MPD threshold are encountered, are the fundamental muscle activity profiles and kinematics of the ankle altered?

6.2 Methods

6.2.1 Participants

This study utilised a within-subject experimental design. Following informed consent participants attended one testing session at the Motion Analysis Laboratory, Curtin University. Eight healthy participants (3 male, 5 female; mean (SD) age 31 (2), height 172.4 (9.1) cm, body mass 72.8 (15.7) kg) free of any pain or functional limitations were tested. All participants were right foot dominant.

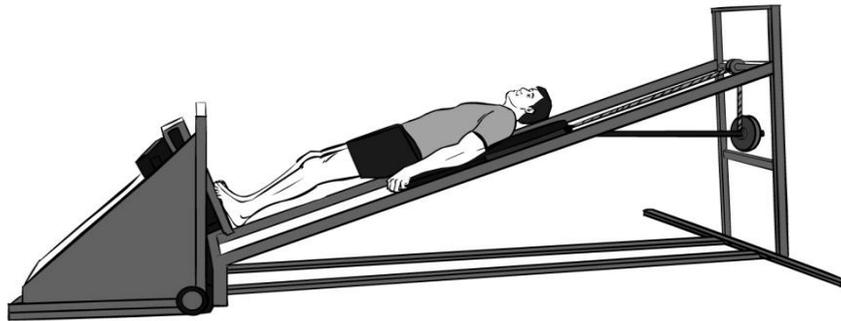


Figure 6.1 Double leg hopping on the custom built sleigh apparatus

The sleigh was set to 20 degrees inclination with a mass offset of the low friction sled – reproduced from (Gibson, Campbell et al. 2013)

6.2.2 Procedure

The participants were requested to perform multiple hopping trials under two conditions - Perceived Hopping (P) and Subperceptual Hopping (SubP). Both P and SubP entailed three trials of 5 continuous bilateral hops at the participants' preferred frequency and natural ground contact time. A sliding floor mechanism (Figure 6.2) allowed the researchers to adjust the height of the landing surface during the flight phase of a hop. One predetermined surface height change was made in the upwards direction under the right foot (all participants were right foot dominant) on either the second, third or fourth hop during each trial. The randomised floor height changes were of either 6mm for SubP trials or 36mm for the P trials. These increments were determined from piloting and represented surface height changes that participants would likely not perceive or would perceive respectively based on previously published data (Travers, Debenham et al. 2013). Participants were instructed in advance that an upwards change in surface height would occur during each trial, but they may or may not perceive it. They were instructed to shout 'change' to notify the testers if they perceived the change. This protocol has been published previously and has demonstrated an ICC of up to .774 for within day testing indicating strong reliability (Travers, Debenham et al. 2013). Importantly, the choice of the SubP and P perturbations was based on pilot data which was validated by 100% perception of the perturbation for the

P condition and 100% non-perception of the perturbation for the SubP condition for the entire cohort of the present study.

All hopping was performed whilst minimising knee flexion (no external fixation was used), and participants kept their eyes closed and were wearing headphones with background music to eliminate auditory feedback.

6.2.3 Familiarisation

A ten minute familiarisation period was conducted for all participants to become comfortable with both hopping conditions and the use of the sleigh. In particular, familiarisation was performed with both eyes closed (replicating the test proper) and eyes open so that participants could be sure that floor height changes were real and of little risk, irrespective of perception.



Figure 6.2 The Sliding Floor Mechanism

This image shows a difference in landing surface height of 36mm between the two feet (Dominant Bilateral Up)

6.2.4 3-D Motion Analysis & EMG

3-D Motion Analysis and EMG data was recorded, handled and analysed as described for Chapter 5 - Motor strategy modulation when expecting an external perturbation – a protective motor response or a searching sensory strategy?

6.2.5 Kinematic Variables

Sensory feedback during dynamic tasks has been related to the stimulus of the muscle fibres and receptors of the passive structures (Jami 1992; Refshauge, Chan et al. 1995; Wise, Gregory et al. 1996). Both the amplitude and rate of loading have been shown to impact on the afferent feedback thresholds (Hall and McCloskey 1983; Wise, Gregory et al. 1998; Salles, Alves et al. 2011). Furthermore, both variables are relevant to the measurement of stiffness (Hobara, Kanosue et al. 2007). As a result the amplitude and rate of the stretch during loading was defined for each individual hop.

As per Chapter 3 – Sleigh vs Upright, the following kinematic variables were outputted:

5. Ankle angle at contact in sagittal plane (θ_{contact})
6. Peak dorsiflexion angle (θ_{peak})
7. Stretch Amplitude ($\theta_{\text{c-p}}$) in degrees ($^{\circ}$): the change in the ankle joint angle from landing (contact) to the most dorsiflexed point
8. Ankle angle at take-off θ in sagittal plane ($\theta_{\text{take-off}}$)
9. Stretch velocity (θ_{vel}) in degrees/ second ($^{\circ}/\text{s}$): the mean dorsiflexion angular velocity, equating to the ratio of the dorsiflexion amplitude to the time interval from landing to the most dorsiflexed point.

6.2.6 Analysis

All data was temporally synchronised and recorded to hard disc on the Motion Laboratory dedicated hardware running a customised Labview program (National Instruments Inc). We analysed the data from the hop on the changed floor height from each of the trials, giving 3 hops for the P condition and 3 hops for the SubP condition per participant. Kinematic data was analysed for eight participants and we had EMG data for six participants due to signal loss. All Data analysis was performed using statistical software (IBM SPSS Statistics version 20: IBM Corp©). An alpha of 0.05 was used to represent statistical significance for all comparisons. A series of paired samples t-tests was utilized to determine significant differences in mean values for the right leg (side of change) across the two conditions for all variables.

6.3 Results

6.3.1 Kinematic Data

The participants demonstrated no statistically significant differences in contact duration, flight duration, hopping frequency or stiffness (all $p > .05$) when compared across conditions (See Table 6.1). As per Table 6.2 the only statistically significant kinematic change observed was a mean increase in peak dorsiflexion of 2.7 degrees ($p = .028$; 95% CI lower bound 0.4; 95% CI upper 5.0). Finally, Table 6.3 illustrates the comparison of stretch amplitude and stretch velocity between SubP and P.

Table 6.1 Differences in the descriptive variables between SubP and P

Variable	SubP mean (SD)	P mean (SD)	Mean Diff	95% CI	p
Hopping Frequency	1.31 (.13)	1.23 (.15)	.08	-.03 to .19	.117
Contact Duration (ms)	361 (42)	383(40)	-23	-66 to 21	.263
Flight Duration (ms)	409 (75)	458 (101)	-49	-134 to 35	.211
Stiffness (kN.m-1)	9.96 (3.41)	8.73 (2.18)	1.23	-.90 to 3.35	.215

*Denotes a significant increase between SubP and P conditions; Mean Diff = Difference in Means; 95% CI = 95% Confidence Interval of the difference

Table 6.2 Sagittal plane ankle angle (in degrees) at specific time points during the contact phase for SubP and P

Time Point	SubP mean (SD)	P mean (SD)	Mean Diff	95% CI	p
θ_{contact}	-23.1 (4.5)	-21.7 (6.7)	1.35	-4.3 to 1.6	.318
θ_{peak}	3.8 (5.6)	6.5 (4.5)*	2.7	0.4 to 5.0	.028
$\theta_{\text{take-off}}$	-34.9 (6.1)	30.9 (8.8)	4.1	-9.6 to 1.5	.125

*Denotes a significant increase between SubP and P conditions; Mean Diff = Difference in Means; 95% CI = 95% Confidence Interval of the difference; Minus values denote plantarflexion

Table 6.3 Differences in Stretch Amplitude and Stretch Velocity between SubP and P conditions

Variable	SubP mean (SD)	P mean (SD)	Mean Diff	95% CI	p
θ_{c-p} (°)	26.9 (4.0)	28.2 (4.0)	1.34	-3.8 to 1.1	.231
θ_{vel} (°/s)	348.3 (58.1)	357.0 (53.7)	8.7	-28.8 to 11.4	.342

*Denotes a significant increase between SubP and P conditions; Mean Diff = Difference in Means; 95% CI = 95% Confidence Interval of the difference

Figure 6.3 illustrates the time normalised saggital plane ankle profiles during the SubP and P conditions. Notably, no significant differences were between conditions for any period of the contact phase.

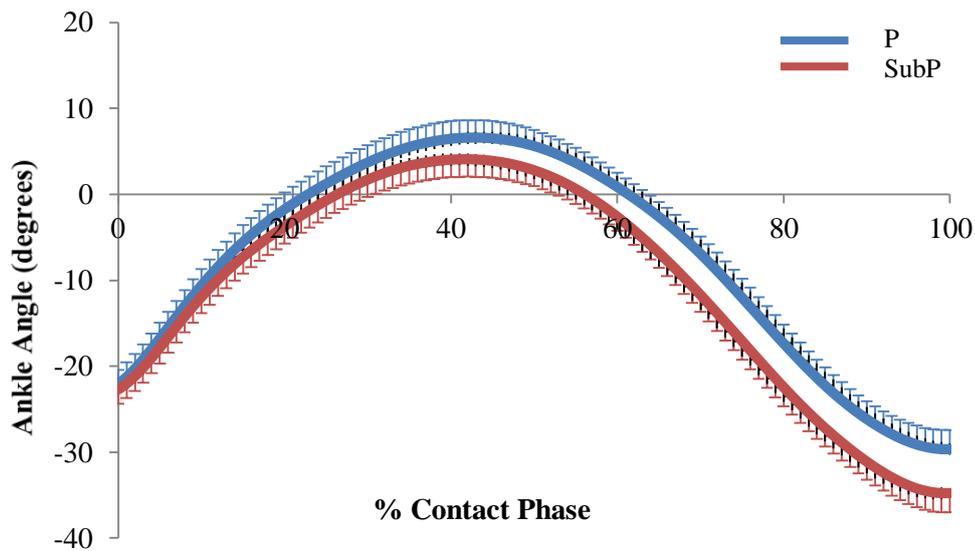


Figure 6.3 Time normalised saggital plane ankle profile during the SubP condition and the P condition

All graphs have been normalised to contact duration where 0 = contact, 100 = toe-off / end of the contact phase (X-Axis). Y-Axis: ankle angle; positive values = Dorsiflexion; Minus values = Plantarflexion; Error bars represent standard error (n = 8)

6.3.2 EMG Data

Table 4 outlines the EMG signal onsets and time to peak (in ms relative to contact) for SubP and P conditions. No significant differences were observed for any of these discrete EMG variables between conditions (all $p > .142$).

Table 6.4 EMG onsets and time to peak (in ms relative to contact) for SubP and P conditions

	SubP	P	
	Mean (SD)	Mean (SD)	p value
	95% CI	95% CI	
Soleus			
Onset	37 (52)	73 (24)	.142
	4 to 79	53 to 92	
Time to Peak	110 (106)	192 (30)	.177
	85 to 195	168 to 217	
Medial Gastrocnemius			
Onset	5 (70)	14 (25)	.480
	-61 to 50	-7 to 33	
Time to Peak	157 (41)	182 (18)	.358
	123 to 190	167 to 196	
Tibialis Anterior			
Onset	42 (21)	35 (51)	.971
	25 to 59	5 to 76	
Time to Peak	145 (12)	168 (85)	.522
	46 to 244	99 to 234	

95% CI = 95% Confidence Interval of the mean; Negative values refer to duration prior contact; SD = Standard Deviation

Figure 6.4 illustrates the time normalised EMG profiles for the Sol, TibAnt and MG during the contact phase for both conditions and ankles. Sol and MG had their own specific window where the EMG activity significantly increased for the P condition (all $p < .05$).

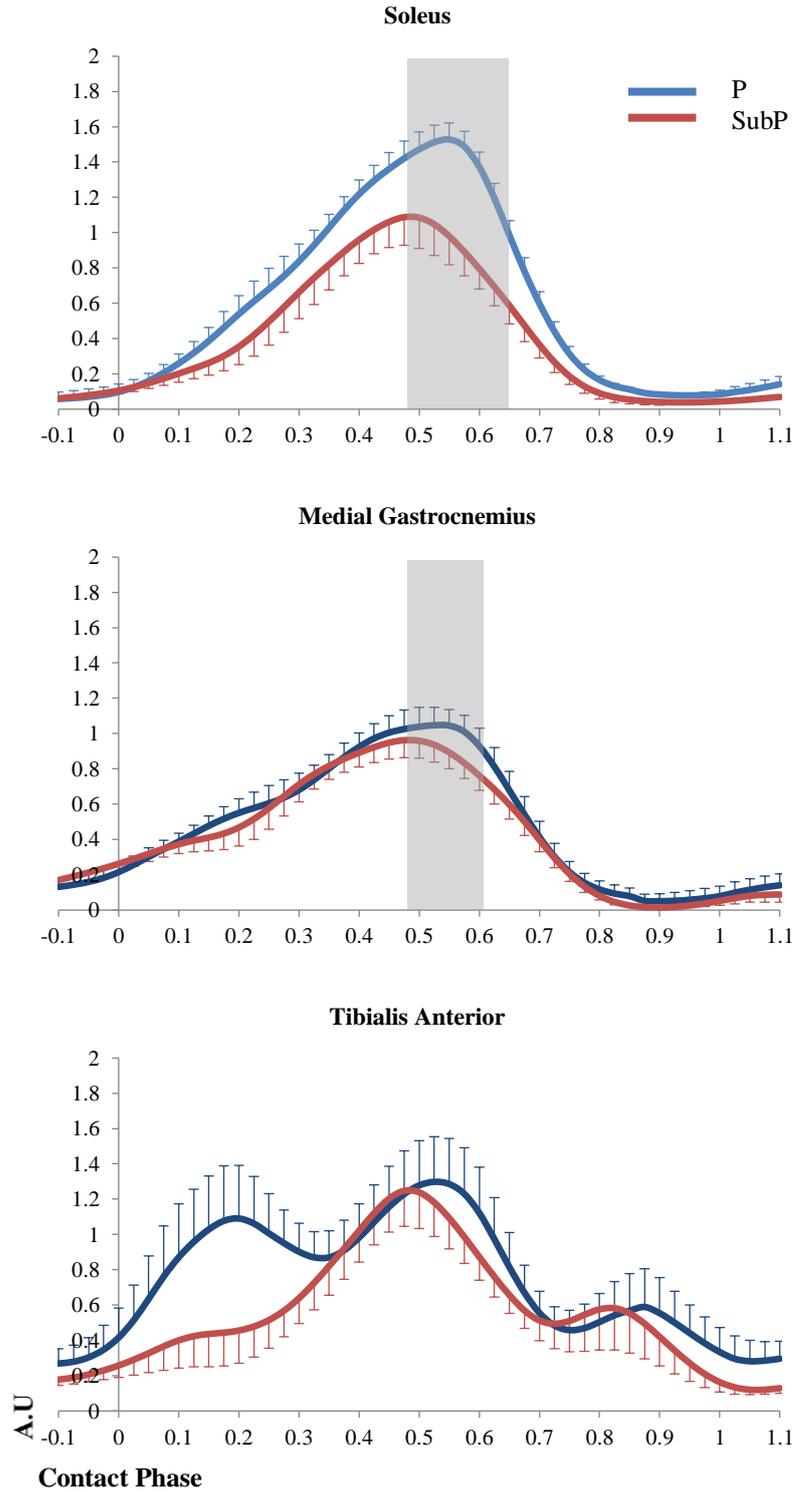


Figure 6.4 Time normalised, group mean linear envelopes for the Sol, TibAnt and MG muscles with standard error (error bars) during the P condition and the SubP condition

All graphs have been normalised relative to the contact phase where 0 = contact, 1 = toe-off / end of the contact phase and minus values indicate pre-contact. X-axis: time (s); Y-Axis: the 1.0 arbitrary unit (A.U) reflects the peak value for the familiarization protocol; Shaded areas indicate ranges where a significant difference existed between conditions; Error bars represent standard error (n = 6).

6.4 Discussion

Testing proprioception relies heavily on one's ability to cognitively perceive changes in a body segment position or orientation. In particular the MPD testing protocol examines participants' ability to cognitively detect small changes in floor height during repeated stretch shortening cycles. It has been demonstrated previously that the mean minimal detectable change in floor height during hopping is approximately 16mm (Travers, Debenham et al. 2013). The present study utilised the MPD experimental paradigm to investigate the effect of cognitive perception of a change in floor height (36mm) on the associated motor outputs as compared to subliminal perturbations (6mm). The authors are unaware of any other study which specifically introduces subperceptual external perturbations and measures the associated changes in behaviour during hopping.

We observed no significant changes in contact duration, flight time, stiffness or hopping frequency (all $p > .05$) between the SubP and P conditions. Furthermore, we observed no statistically significant differences in ankle θ_{contact} , $\theta_{\text{c-p}}$ or θ_{vel} when compared across conditions (all $p > .05$). The only kinematic variable that was statistically different across conditions was the θ_{peak} which increased from 3.8 (+/- 5.6) degrees for Sub P to 6.5 (+/- 4.5) for P ($p = .028$). It could be tempting to speculate that perception of the perturbation may be linked to the mean increase of 2.7 (95% CI 0.4 to 5.0) degrees dorsiflexion we observed for the P trials. However, this is a relatively small difference in range when considered against the entire stretch amplitude of 26.9 (+/-4.0) degrees and 28.2 (+/-4.0) degrees for the SubP and P conditions respectively. It is plausible that the increase in θ_{peak} observed for the P condition was a direct result of the larger (36mm) floor height increase introduced during these trials. As the participants were hopping bilaterally it is possible that the increase in peak dorsiflexion under the right / perturbed foot may be due to the discrepancy in floor height between the two feet making it difficult to attribute this to cognitive perception of the perturbation.

In a previous study we observed a significant increase ($p < .020$) in stretch velocity which was attributable to the participants' expectation of an impending change in floor height (Travers et al 2013, See also Chapter 5). It was unclear whether such an increase in stretch velocity represented a strategy to increase leg stiffness or an attempt to optimise proprioceptive sensitivity via the stretch-velocity relationship (Ribot-Ciscar and Roll 1998; Proske, Wise et al. 2000). We concluded that further research was required to investigate the relationship between stretch velocity and successful perception of surface height changes

during the MPD test (Travers et al 2013, Unpublished / Chapter 3). In the present study, the stretch velocity was 348.3 (+/- 58.1) °/s and 357.0 (+/- 53.7) °/s for the Sub P and P conditions respectively. We observed no significant difference in stretch velocity ($p = .342$) whether participants successfully perceived the change in floor height or whether the change was below their threshold of detection. In the existing proprioception literature it is accepted that the rate of loading heavily influences detection of movement (Hall and McCloskey 1983; Wise, Gregory et al. 1998; Salles, Alves et al. 2011). For example, Hall and McCloskey (1983) observed that the detection of threshold of the forearm can be increased up to eightfold with reduced speed of displacement (stretch velocity). Therefore it could be expected that in the presence of a difference in stretch velocity participants could have increased acuity during the MPD test. However, it appears from our findings that ankle stretch velocity is not related to detection of changes in floor height during repeated SSC.

Figure 6.4 illustrates the time normalised EMG profiles for the Sol, TibAnt and MG during the contact phase for both conditions. Sol and MG had their own specific window where the EMG activity significantly increased for the P condition ($p < .05$). For Sol this window commenced at 47.5% and finished at 65% of the duration of the contact phase. For MG the window in question extended from 47.5% to 60% of the duration of the contact phase. Both of these windows occurred just after the period of peak DF illustrated in Figure 6.3 at 43% of the contact phase duration for the P condition. It is unclear whether these EMG changes to the plantarflexor muscles are factors which trigger cognitive perception or whether they are a behavioural change resulting from detection of the larger (36mm) increase in floor height.

As discussed above, the only mechanical change observed between the two conditions was a small change in θ_{peak} after the introduction of the 36mm block. It is a limitation of the present study that relative loading of the two limbs was not documented. It is possible that raising the floor by 36mm may have increased the loading of the perturbed ankle leading to the observed increase in dorsiflexion and this may be a possible driver of detection. It may similarly be hypothesised then that the observed increase in plantarflexor activity was a direct result of this increase in load.

6.5 Conclusion

Perception of an expected perturbation during hopping did not manifest in major behavioural changes as represented by kinematic and EMG measurement in the present cohort. All participants perceived the 36mm increase in floor height. A small increase in peak dorsiflexion and a specific window of increased activity of the plantarflexor muscles near the point of peak dorsiflexion for the 36mm perturbation were observed. However, it is unclear if these small changes in kinematics and muscle activity represent drivers of perception or whether they are the result of the detection of the mechanical change under the foot.

7. Perceived changes in floor height occur without detectable changes in range

7.1 Introduction

Traditional tests of proprioceptive acuity involving position matching, force matching and detection of passive movements represent convenient laboratory based tests, but do not replicate normal function (Proske, Wise et al. 2000). Researchers have attempted to use more functionally relevant tests by assessing sagittal plane, ankle joint position matching and movement detection sense in standing (Blaszczyk, Hansen et al. 1993; Blaszczyk, Lowe et al. 1993; Refshauge and Fitzpatrick 1995). The most commonly utilised global proprioception test in the clinical setting is the Romberg Test which measures the increase in body sway when standing patients close their eyes (Proske and Gandevia 2012). Whereas isolating lower-limb proprioception clinically is often determined via comparing the movement detection threshold of the big toes (Proske and Gandevia 2012). However, it is difficult to utilise this measure at the big toe to quantify changes at the ankle or elsewhere in the leg due to anatomical limitations (Refshauge, Chan et al. 1995). Furthermore, there is a paucity of research examining proprioception during more dynamic task such as locomotion.

Recently the Minimal Perceptible Difference (MPD) test has been presented as a measure proprioceptive acuity during self-paced active muscle function of the lower limb with a specific focus on the ankle (Travers, Debenham et al. 2013). In keeping with the central comparison between sensory and motor signals model (Bays and Wolpert 2007), the MPD test assesses the ability to detect changes in the external environment as opposed to other ankle proprioceptive testings that isolate position matching or joint movement detection tests (see Chapter 1). Therefore, a key consideration is the effect of participants' expectation of the introduced height change during the MPD test. It has been widely demonstrated that participants modify their motor behaviour via feedforward mechanisms when expecting an impending perturbation (Moritz and Farley 2004; Taube, Leukel et al. 2012). This is especially relevant to the MPD protocol as the participants were specifically instructed to search for the change in floor height during testing rather than just passively experience them. When performing the MPD test participants demonstrated a change in motor behaviour from their baseline hopping (BH) strategy when they expected and were instructed to search for an impending change in floor height (Chapter 5). This expectation hopping (EH) pattern was characterised by a significant increase in the stretch velocity and leg stiffness bilaterally. Furthermore, there were increases in muscle activity in windows that appear specific to either the uni-articular and mono-articular muscles at the ankle bilaterally.

Therefore, it is clear that the expectation of a safe, subtle and familiar perturbation is sufficient to induce significant kinematic and muscle activity changes to the baseline hopping strategy when performing the MPD test.

Having established EH as the default performance during the MPD test, we examined whether the motor responses to the introduced change in surface height differed depending on the participants' cognitive detection of the perturbation (Chapter 6). This was achieved by comparing the kinematic and EMG data from the perturbed (dominant) ankle across Subperceptual (SubP) and Perceived (P) conditions. SubP entailed a 6mm floor height change whereas P entailed a 36mm floor height change during the hopping trial. The two conditions were derived from pilot work developing the MPD where it was observed that the 6mm and 36mm height changes corresponded to the limits of threshold of perceptive range for normal subjects. Importantly, no change in the kinematic profile of the ankle when compared across conditions was observed (Chapter 6). However, the plantarflexor group showed an increase in muscle activity around the period of peak physiological loading for the P condition, but it is not clear whether this represented a driver of detection, a result of the cognitive perception or a combination of both factors. It is evident that the motor responses to a detected perturbation differed little from the responses to a subliminal perturbation (Chapter 6). It is also clear that cueing participants to search for a change in floor height during the MPD test resulted in overt changes to their baseline hopping strategy (BH) (Chapter 5). The new hopping strategy observed when expecting a perturbation was labelled expectation hopping (EH). It is possible that the observed changes in motor performance may represent a feedforward response to the pending perturbation, or the EH strategy may represent a searching strategy in order to detect the safe, familiar and subtle perturbation that was expected. Therefore, this chapter aims to investigate any deviation from the EH condition occurring as a result of detected or undetected perturbations. To this end the authors have utilised the MPD test to answer the following questions:

1. Do the sagittal plane kinematic profiles of the perturbed ankle for the SubP and P conditions differ from EH?
2. Do the muscle activity profiles of the perturbed ankle for the SubP and P conditions differ from EH?

7.2 Methods

7.2.1 Analysis

For the present study, the kinematic and EMG data from the EH condition was considered the baseline hopping performance. This was then compared to the outputs from the SubP and P conditions. Kinematic data was analysed for eight participants and we had EMG data for six participants do due signal loss. All data analysis was performed using statistical software (IBM SPSS Statistics version 20: IBM Corp©). An alpha of 0.05 was used to represent statistical significance for all comparisons. A series of paired samples t-tests was utilized to determine significant differences in mean values for the right leg (side of change) for all comparisons.

7.3 Results

7.3.1 Kinematic Data

The participants demonstrated no statistically significant differences in contact duration, flight duration, or stiffness (all $p > .05$) when compared across conditions (See Table 7.1). The only statistically significant kinematic change observed was a mean decrease in hopping frequency of .09Hz for the P condition ($p = .049$; 95% CI lower bound -.18; 95% CI upper -.01) compared to EH.

Table 7.1 Differences in the descriptive variables comparing EH with SubP & P

Variable	EH mean (SD)	Comparison Condition	Mean (SD)	Mean Diff	95% CI	p
Hopping Frequency	1.32 (.09)	SubP	1.31 (.13)	-.01	-.07 to .05	.669
		P	1.23 (.15)*	-.09	-.18 to -.01	.049
Contact Duration (ms)	356 (30)	SubP	361 (42)	5	-10 to 21	.481
		P	383 (40)	27	-.11 to .66	.131
Flight Duration (ms)	427 (67)	SubP	409 (75)	-18	-63 to 126	.461
		P	458 (101)	31	-136 to 83	.541
Stiffness (kN.m-1)	10.21 (3.07)	SubP	9.96 (3.41)	-.25	-3.40 to .45	.112
		P	8.73 (2.18)	-1.47	-.77 to .27	.291

*Denotes a significant difference between EH and the comparison conditions; Mean Diff = Difference in Means; 95% CI = 95% Confidence Interval of the difference; SD = Standard Deviation

Table 7.2 contains the differences in the sagittal plane ankle profile comparing EH with SubP & P conditions. No significant differences were observed for any of these comparisons.

Table 7.2 Differences in the sagittal plane ankle profile comparing EH with SubP & P

Variable	EH mean (SD)	Comparison Condition	Mean (SD)	Mean Diff	95% CI	p
Contact	-23.0 (5.4)	SubP	-23.1 (4.5)	.01	-1.20 to 1.03	.190
		P	-21.7 (6.7)	-1.29	-.80 to 3.33	.705
Peak Dorsiflexion	4.7 (4.6)	SubP	3.8 (5.6)	-.92	-2.94 to 1.10	.314
		P	6.5 (4.5)	1.77	-.22 to 3.75	.073
Take-off	34.4 (6.6)	SubP	34.9 (6.1)	.50	-2.33 to 1.35	.545
		P	30.9 (8.8)	-3.58	-.30 to 7.46	.065

*Denotes a significant difference between EH and the comparison conditions; Mean Diff = Difference in Means; 95% CI = 95% Confidence Interval of the difference; Negative values refer to Dorsiflexion angle; SD = Standard Deviation

Finally, Table 7.3 illustrates the comparison of stretch amplitude and stretch velocity between EH and SubP, EH and P. No significant differences were observed for any of these comparisons.

Table 7.3 Differences in stretch amplitude and stretch velocity comparing EH with SubP & P

Variable	EH mean (SD)	Comparison Condition	Mean (SD)	Mean Diff	95% CI	p
Stretch Amplitude (degrees)	27.7 (4.5)	SubP	26.9 (4.0)	-.80	-3.54 to 1.86	.488
		P	28.2 (4.0)	.50	-.92 to 1.92	.431
Stretch Velocity (°/s)	344.2 (48.3)	SubP	348.3 (58.1)	4.1	-26.4 to 34.7	.758
		P	357.0 (53.7)	12.8	-12.2 to 37.8	.265

*Denotes a significant difference between EH and the comparison conditions; Mean Diff = Difference in Means; 95% CI = 95% Confidence Interval of the difference

7.3.2 EMG Data

Table 7.4 outlines the EMG onsets and time to peak (in ms relative to contact) for EH, SubP and P. No significant differences were observed for any of these discrete EMG variables between conditions (all $p > .609$).

Table 7.4 EMG onsets and time to peak (in ms relative to contact) for EH vs. SubP & P

	EH vs. SubP			EH vs. P		
	EH	SubP	p	EH	P	p
	Mean (SD) 95% CI	Mean (SD) 95% CI		Mean (SD) 95% CI	Mean (SD) 95% CI	
Soleus						
Onset	77 (33) 50 to 103	37 (52) 4 to 79	.146	77 (33) 50 to 103	73 (24) 53 to 92	.613
Time to Peak	114 (19) 101 to 125	110 (106) 85 to 195	.212	114 (19) 101 to 125	192 (30) 168 to 217	.798
Medial Gastrocnemius						
Onset	20 (45) -15 to 57	5 (70) -61 to 50	.219	20 (45) -15 to 57	14 (25) -7 to 33	.609
Time to Peak	176 (39) 145 to 208	157 (41) 123 to 190	.439	176 (39) 145 to 208	182 (18) 167 to 196	.667
Tibialis Anterior						
Onset	36 (31) 9 to 59	42 (21) 25 to 59	.815	36 (31) 9 to 59	35 (51) 5 to 76	.978
Time to Peak	175 (48) 137 to 213	145 (12) 46 to 244	.727	175 (48) 137 to 213	168 (85) 99 to 234	.840

*Denotes a significant difference between EH and the comparison conditions; Mean Diff = Difference in Means; 95% CI = 95% Confidence Interval of the difference; SD = Standard Deviation

Finally, Figure 7.1 illustrates the time normalised EMG profiles for the Sol, TibAnt and MG during the contact phase for all three conditions. Importantly, the only statistically significant difference in the EMG profiles detected was an increase in medial gastrocnemius activity in a window from 55% of the contact phase to 62.5% of the contact phase for P when compared to EH (all $p < .05$).

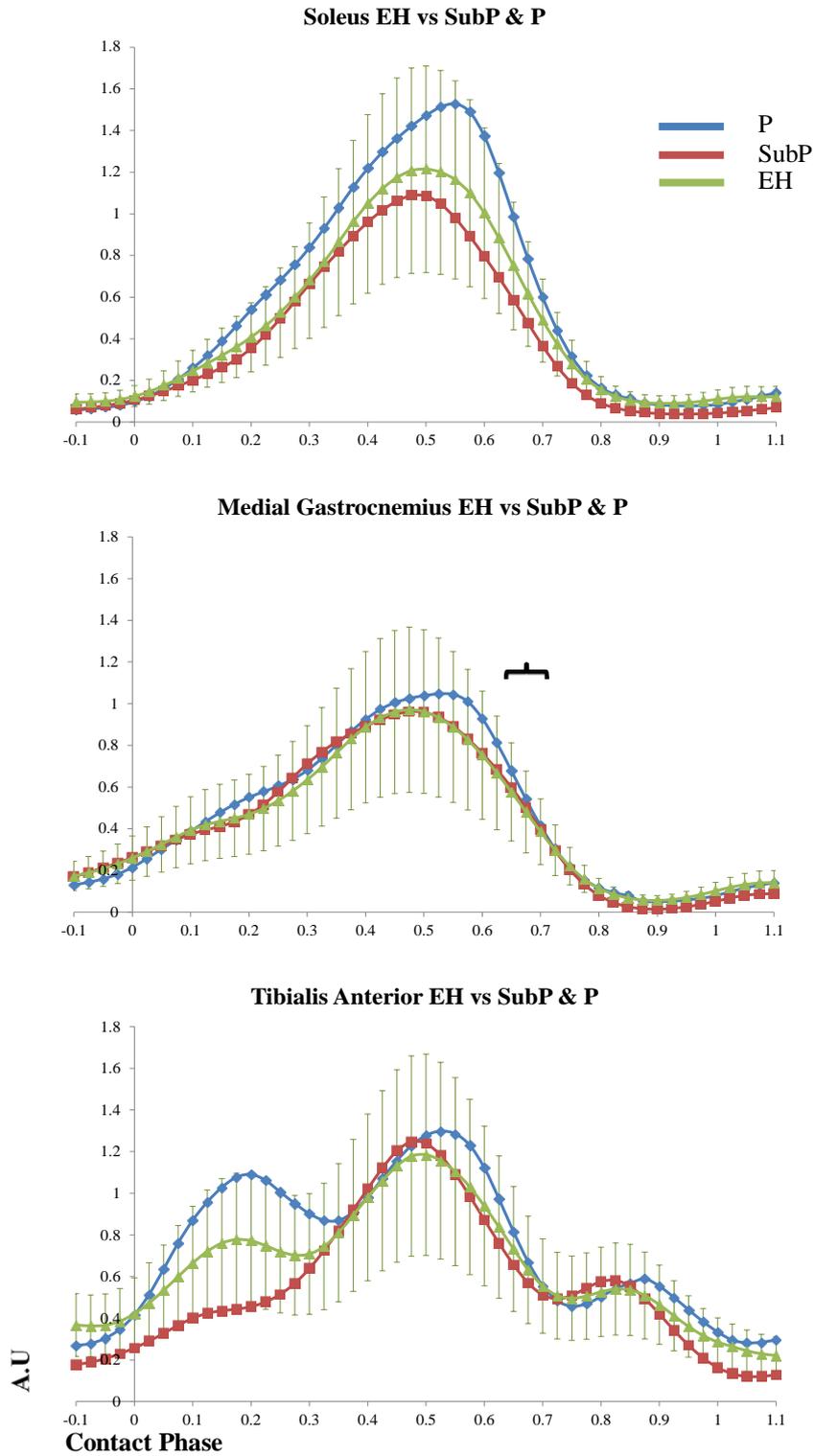


Figure 7.1 Time normalised, group mean linear envelopes for the Sol, TibAnt and MG muscles during the EH (green line), P (blue line) and SubP (red line) conditions

Note: the error bars represent the standard error for the EH condition (n = 6). All graphs have been normalised relative to the contact phase where 0 = contact, 1 = toe-off / end of the contact phase and minus values indicate pre-contact. X-axis: time (s); Y-Axis: the 1.0 arbitrary unit (A.U) reflects the peak value for the familiarization protocol; The brackets indicate any time points where a statistically significant difference in the EMG signal was detected when compared to EH (all $p < .05$)

7.4 Discussion

Proprioceptive testing represents an attempt to quantify one's ability to cognitively perceive changes in a body segment's position or orientation. The MPD testing protocol examines participants' ability to cognitively detect small changes in the external environment such as landing surface height during repeated stretch shortening cycles. This research represents a shift towards investigating proprioception in a task which is dynamic, repeated, weight bearing and more closely resembles real world activities such as gait. In the present analysis we have examined whether the motor patterns observed for the SubP and P conditions differ from the default hopping strategy during the MPD test. The following sections will discuss the findings of the present analysis in the context of the wider thesis and then in the context of other research.

7.4.1 Kinematic Data

The only statistically significant difference observed for any of our kinematic derived variables in the present analysis was a decrease in mean hopping frequency of .09Hz for the P condition when compared to EH. It is intuitive that the introduction of a 36mm increase in floor height (30mm difference from the comparative SubP) would result in an earlier foot contact and thus reduce the flight time of the hopping performance. Such a reduction in flight time would result in an increase in hopping frequency if the contact time remained the same. However, we have observed a slight increase in both contact time (mean increase 27 ms; $p = .131$) and flight time (mean increase 31 ms; $p = .541$) for the P condition when compared to EH. The combined effect of these increases in contact and flight times resulted in a mean decrease of the hopping frequency for P. Interestingly, in Chapter 5 we were unable to detect a significant difference in hopping frequency between the SubP and P conditions ($p = .117$). It is possible that although the difference in hopping frequency observed here between EH and P achieved statistical significance, it may not carry practical significance. The absolute difference in hopping frequency between EH and P observed here was 0.09Hz and may appear small. However it actually equals the standard deviation in hopping frequency for the EH condition giving an effect size of 1.0. Therefore, it would be an oversight to dismiss this 'small absolute difference' and we cannot negate this temporal difference in hopping performance as a possible driver of detection for the P condition.

We have observed no statistically significant changes in contact duration, flight time or stiffness (all $p > .05$) between EH and either the SubP or P conditions. Furthermore, we

observed no significant differences in ankle angle at contact, peak dorsiflexion, ankle angle at take-off, stretch amplitude nor stretch velocity when compared between EH and either the SubP or P conditions (all $p > .05$). Thus, we are confident that the kinematic outputs associated with both SubP and P are not statistically significantly different from those for EH. The significance of these findings is that our participants were hopping in a similar manner irrespective of the impending perturbation, irrespective of the magnitude of the perturbation (6mm vs. 36mm) and irrespective of perception of the perturbation during testing.

The current study and the wider thesis have focused on the ankle region however; it is possible that perception of perturbations may be driven by mechanical cues more proximally in the leg. However, there was a lack of demonstrable differences in the kinematic variables assessed in the present study. Therefore, it seems these factors may not determine perception during the MPD test. Furthermore it appears that there is a large degree of redundancy in the mechanics of the ankle during dynamic tasks. This is significant as a key construct of the traditional proprioception research paradigm is to link perception of range related measures (position matching, detection of movement) to proprioceptive acuity (Blaszczyk, Hansen et al. 1993; Blaszczyk, Lowe et al. 1993; Matre, Arendt-Neilsen et al. 2002; Lund, Juul-Kristensen et al. 2008; Lowrey, Strzalkowski et al. 2010; Salles, Costa et al. 2011) yet we have been unable to link detection of changes in the external environment to any of our derived variables that are relevant to ankle position or movement e.g. ankle range, stretch amplitude, stretch velocity or stiffness.

Previous studies have reported open chain repositioning errors and movement detection errors of the ankle of less than 3 degrees in the sagittal plane for healthy populations (Matre, Arendt-Neilsen et al. 2002; Fu and Hui-Chan 2007). This suggests that individuals are extremely sensitive at cognitively perceiving changes in ankle joint position and movement in controlled experimental settings. One might consider this precise acuity may drive detection of ankle perturbations in more functional and dynamic scenarios such as bilateral hopping. It is an interesting observation that the introduction of neither subliminal nor perceived perturbations resulted in minimal kinematic and muscle activity profile changes at the ankle during the MPD test. Importantly, we observed a 100% detection rate for the P condition and yet no statistically significant differences were observed for any of our range related measures for the ankle including ankle angle at contact, peak dorsiflexion, stretch amplitude and stretch velocity (all $p > .05$). This raises questions for future research

regarding what factors actually drive detection during the MPD test. More research is needed to establish the functional significance of the purported finely-tuned proprioceptive acuity associated with open chain proprioceptive measures for the ankle. Furthermore, an important evolution of this thesis line would be to compare individuals' performance in traditional open chain proprioceptive tests of the ankle and sensitivity to the MPD test.

The closest methodology to the MPD test utilised a closed chain, standing position matching test of the ankle (Blaszczyk, Hansen et al. 1993). Their healthy cohort stood with one foot on level ground and the other foot on a sagittal plane tilt platform. Following passive tilting of that ankle they were requested to actively return it to the reference position matching the grounded ankle (Blaszczyk, Hansen et al. 1993). Interestingly they observed matching errors of less than 1 degree in both plantarflexion and dorsiflexion and suggested that the high degree of accuracy in their test maybe due to the perception of gravitationally induced forces resulting from the upright position (Blaszczyk, Hansen et al. 1993). This is a reasonable conclusion and is substantiated in part by other research which has demonstrated that muscle spindle signals and sense of effort combine to optimise sense of limb position (Winter, Allen et al. 2005). However, it is plausible that the high degree of accuracy observed in standing tests of ankle proprioception is contributed to by a bilateral comparison of both weight bearing limbs. This hypothesis is consistent with increased sensitivity during bilateral hopping performing the MPD test when compared to alternate hopping (Travers, Debenham et al. 2013).

An underlying assumption in the wider literature is that better proprioception performance is displayed by greater sensitivity to movement (or position) and that this is a continuum making it preferable to be more sensitive during proprioceptive testing. This may be true when considering upright stance and maintenance of posture where it is clear that healthy people exhibit extremely high sensitivity to sagittal plane ankle position during upright stance (Blaszczyk, Hansen et al. 1993). It is consistent therefore, that acuity declines with aging along with other features considered integral to maintenance of upright posture such as maximum voluntary excursion of the centre of mass and postural sway (Blaszczyk, Hansen et al. 1993; Blaszczyk, Lowe et al. 1993; Blaszczyk, Lowe et al. 1994; McClenaghan, Williams et al. 1995). However, it has been demonstrated that healthy people operate with significant proprioceptive redundancy when performing the MPD test - the mean minimal detectable change in floor height during hopping being measured at approximately 16mm and 27mm for bilateral and alternating hopping respectively (Travers, Debenham et al.

2013). It has been postulated previously that this may reflect the specific proprioceptive requirements of the different tasks of maintaining upright posture and maintaining centre of mass dynamic during locomotion. The absence of significant differences in the sagittal plane kinematic outputs of the perturbed ankle when compared to adaptive expectation changes in the current analysis may add further weight to this argument. It is plausible that during more dynamic tasks like repeated SSC, detection of changes in the foot-floor interface may not be actually driven by detection of differences in range or position between both ankles. Thus, the high acuity observed in standing proprioceptive tests of the ankle may be redundant when participants' intention changes from maintaining upright posture to maintaining centre of mass dynamics. Once again, further research is necessary to establish any correlation between performances during such closed chain testing methods to the more dynamic MPD test.

7.4.2 EMG Data

The findings from the EMG analysis further substantiate the similarities across the conditions. Figure 7.1 illustrates the time normalised EMG profiles for the Sol, TibAnt and MG during the contact phase for all three conditions. The only statistically significant difference observed was a short window where MG activity significantly increased for the P condition ($p < .05$). This window extended from 55% of the contact phase to 62.5% of the contact phase when compared to EH. This finding bears some similarity to Chapter 6 where we observed an increase ($p > .05$) in MG activity from 47.5% to 60% of the duration of the contact phase for the P condition compared to SubP which is consistent with the window of increased MG activity in the current analysis. However, in the Chapter 6 we also observed a simultaneous increase in Sol activity for the P condition compared to SubP which has not been observed between EH and P (all $p > .05$). It is evident that the introduction of a subliminal or perceived perturbation resulted in minimal muscle activity profile changes in the Sol, MG and TibAnt muscles when compared to EH. Furthermore, despite the consistent finding of increased MG activity following the introduction of a perceived perturbation in both analyses it remains unclear whether this contributes to cognitive perception or represents a behavioural change resulting from detection of the larger (36mm) increase in floor height.

7.5 Conclusion

Traditional proprioceptive testing methods specifically utilise measures of range or position to quantify proprioceptive function. Importantly, we observed no significant differences in

range or position related outputs for the ankle when either subliminal or perceived perturbations were introduced. These findings question the utility of traditional testing methods of measuring proprioception in the context of normal dynamic function at the ankle. It is clear that further research is required to determine what factors drive perception of changes in floor height during the MPD test.

8. Discussion and summary of major findings

8.1 Introduction

Traditional proprioceptive testing attempts to quantify participants' ability to cognitively perceive changes in body segment position or configuration and has been largely based on very controlled testing of isolated joints (Proske and Gandevia 2012). The present thesis has endeavoured to explore the concept of proprioception during repeated SSC and represents a novel addition to the current body of knowledge. The studies contained in this manuscript have aimed to fill some of the gaps in the literature by merging the experimental constructs commonly used to separately investigate proprioception and dynamic modulation of SSC activity.

The novelty of this research required the development of a new testing paradigm, the MPD test which was administered in a low load environment. Specifically, the MPD testing protocol examines participants' ability to cognitively detect small changes in the landing surface height during sleigh hopping. This research represents a shift towards investigating proprioception in a task which is dynamic, repeated, weight bearing and more closely resembles the normal function of the lower limb (Travers, Debenham et al. 2013). The previous sections discussed the findings of the individual analyses in the context of the wider thesis and also in the context of other research. This chapter shall expand upon and discuss the key findings of the thesis in relation to the existing literature and make suggestions for the future direction of this experimental model.

8.2 A low load environment

The present body of work utilised a sleigh hopping model which was designed to minimise the risk of confounding factors such as balance and fatigue during repeated hopping trials. Though all sleigh systems are different, their use has become a common practice for investigating SSC activity (Flanagan and Harrison 2007; Kramer, Ritzmann et al. 2010; Merritt, Raburn et al. 2012; Furlong and Harrison 2013). It was necessary to examine the reliability of the system and validate some derived variables in order to justify the use of the sleigh system. This section will discuss the pertinent findings substantiating the use of the custom built sleigh apparatus.

A within-subject repeated measures design was utilised to examine leg stiffness and the kinematic profile of the ankle during self-paced, sub-maximal hopping under two conditions – sleigh hopping (SH) and upright hopping (UH). Leg stiffness, θ_{contact} , θ_{peak} , $\theta_{\text{c-p}}$ and $\theta_{\text{take-off}}$ demonstrated excellent reliability on and between both test occasions for SH (all ICC > .975) (Chapter 3). These findings parallel and add to the moderate to strong temporal stability demonstrated for SH (Debenham, Travers et al, Submitted Manuscript - see Appendix 1). Consistent with other researchers, it was concluded that sleigh hopping represents a reliable controlled environment for investigating the SSC (Flanagan and Harrison 2007; Furlong and Harrison 2013). It is necessary to highlight that other studies examining the reliability of sleigh hopping have generated a reliable performance by instructing their participants to maximise their leg stiffness whilst hopping (Kramer, Ritzmann et al. 2010; Furlong and Harrison 2013). Conversely, the participants in the present thesis were specifically instructed to hop at their natural frequency and stiffness. They were instructed to minimise knee movement and thus drive performance from the ankle so it was not surprising to observe that SH is not a replication of UH.

Exploring the relevant literature highlighted that no studies had to date validated estimates of leg stiffness (Dalleau, Belli et al. 2004) during SH. Comparing leg stiffness estimates against the gold standard leg stiffness measure (Cavagna, Franzetti et al. 1988) revealed a very strong correlation ($R^2 = .9534$). Thus, this thesis demonstrated for the first time that the leg stiffness estimate proposed by Dalleau et al (2004) is valid for SH. Furthermore, it had not been demonstrated previously that leg stiffness was attributable to the ankle during SH. By comparing leg stiffness estimates to the gold standard measure for ankle joint stiffness (Farley and Morgenroth 1999) this body of work also demonstrated experimentally that such instruction leads to a SH performance where stiffness is largely attributable to the ankle ($R^2 = .8458$) (Grisbrook et al, Submitted Manuscript – Supplementary Chapter 1). The demonstrated reliability, temporal stability and validity of the pertinent derived variables suggest that the sleigh represents an appropriate environment for exploring the properties of the SSC at the ankle. This was further substantiated by observed correlations for θ_{contact} , $\theta_{\text{take-off}}$, $\theta_{\text{c-p}}$ and leg stiffness between UH and SH (all Pearson correlation .514 - .887) (Chapter 3). These correlations suggest that although UH and SH are different tasks, the same domain is being measured under the present methodology. These combined findings justify the use of the sleigh system for experimentally examining SSC performance throughout this thesis. As such, SH has formed the foundation for the MPD test and also for examining the internal drive to modulate leg stiffness (Unpublished data – Supplementary Chapter 3).

While beyond the scope of this thesis, the above described results suggest that the custom built sleigh apparatus may provide a safe, controlled, reliable and robust tool for examining the SSC performance of the ankle in clinical groups or under experimental pain models. Further research is necessary to examine whether the observed reliability of performance during SH is maintained under multiple loading conditions (i.e. sleigh positioned at different angles or participant externally weighted). As the ankle represents an area of specific interest for this research, SH has not yet been examined without the instruction to constrain knee movement. Therefore it is unknown whether the custom built sleigh apparatus possesses the same utility for examining SSC function further up the limb.

8.3 Validity

A large proportion of the existing literature on the proprioception has been obtained through the study of passive movement detection (Gandevia and McCloskey 1976; Refshauge, Chan et al. 1995; Refshauge and Fitzpatrick 1995; Wise, Gregory et al. 1996; Wise, Gregory et al. 1998; Matre, Arendt-Neilsen et al. 2002) and position matching tasks (Feuerbach, Grabiner et al. 1994; Matre, Arendt-Neilsen et al. 2002; Vuillerme, Chenu et al. 2006; Allen, Ansems et al. 2007; Vuillerme and Cuisinier 2008; Allen, Leung et al. 2010). However, the normal dynamic function of the lower limb entails the performance of repeated SSC (Avela and Komi 1998; Komi 2000; Ogiso, McBride et al. 2002; Horita, Komi et al. 2003; Kallio, Linnamo et al. 2004; Ishikawa, Komi et al. 2006; Ishikawa and Komi 2007) and it has been recognised that proprioception may be a critical component of the sensory input required for successful execution of such movements (Smith, Crawford et al. 2009; Proske and Gandevia 2012). However, there is a paucity of research which examines the proprioceptive function during the SSC. The development of the MPD testing protocol represents an attempt to address the gap in the current literature by investigating participants' ability to detect changes in floor surface height during repeated hopping trials (Travers, Debenham et al. 2013). The reliability of participants to detect these changes has been investigated on a within and between day basis (Chapter 4). The Dominant Bilateral Up condition demonstrated an ICC of up to .774 for within day testing indicating strong reliability and thus was subsequently utilised to examine this experimental paradigm throughout the thesis.

Previous research has highlighted the strategy dependent utility of proprioceptive information by comparing the effect of vibration stimulus on different tasks – upright stance and gait (Ivanenko, Grasso et al. 2000). When performing the MPD test participants appeared to display strategy dependent sensitivity to changes in the floor surface with a

mean MPD score of 15.7mm during bilateral hopping as compared to 26.6mm during alternating foot strikes ($p < .0001$) (Travers, Debenham et al. 2013). Therefore, there is a consistency in observations between these studies. It is likely, that the acuity of these scores was optimised by priming the participants to the side and direction of the pending perturbation as much lower sensitivity (up to 60mm) was observed in the absence of cuing during pilot testing (Travers, Debenham et al. 2013). However, traditional testing methods have demonstrated position matching and movement detection errors of less than 1 degree in sagittal plane movement of the ankle in both sitting and standing (Blaszczyk, Hansen et al. 1993; Refshauge, Chan et al. 1995; Refshauge and Fitzpatrick 1995; Matre, Arendt-Neilsen et al. 2002). The ability of participants to so accurately position match sagittal plane movements of the ankles in standing suggests a finely tuned sensitivity that may be specific to the maintenance of upright stance (Blaszczyk, Hansen et al. 1993). Conversely, the present body of research has observed lower than expected sensitivity to changes in floor height during repeated SSC and a strategy dependency to this sensitivity. Therefore, the findings of this research suggest that humans exhibit a low degree of sensitivity to changes in the foot-floor interface when hopping. This may allow for significant variation in this interface during gait without cognitive perception of the challenge or disturbance of efficient performance and may question the relevance of traditional static proprioceptive tests to more dynamic function (Travers, Debenham et al. 2013).

However, demonstrating reliability of the system and observing consistencies to others' findings (e.g. strategy dependent sensitivity) does not infer validity to the MPD testing paradigm. Comparisons of participants' performance during the MPD test to the established gold standard tests of position matching and movement detection remain necessary. By doing so, it will be possible to establish validity of the MPD testing protocol as a pure test of proprioceptive function during the SSC. Furthermore, such an experiment may help to clarify if we are measuring the same domain during the MPD test. A clear and logical follow on step would be to test the above findings under various loading scenarios. As discussed previously the sleigh provided a safe and controlled environment for the uncontaminated performance of multiple SSC (Unpublished data, Chapter 3, see also Appendix 1 and Supplementary Chapter 2). It is plausible that the low-load environment of the sleigh may have affected the ability of participants to detect changes in surface height whilst hopping. Therefore, an area for future research is to investigate whether altering the angle of the sleigh and thus the loading parameters affects participants' performance of the MPD test.

8.4 Feedforward Control

The feedforward control of SSC activity has been demonstrated extensively in the existing literature (Komi and Gollhofer 1997; Moritz and Farley 2004; Marquez, Aguado et al. 2010; Masahiro and Shingo 2010; Leukel, Taube et al. 2012; Taube, Leukel et al. 2012; Marquez, Aguado et al. 2013). For example, it has been demonstrated that when participants expect a potential “foot in hole scenario” that they adopt a protective strategy via an increase in ankle dorsiflexion at foot strike (van der Linden, Hendricks et al. 2009; Masahiro and Shingo 2010). Despite the more widespread of use sleigh systems to examine SSC no studies existed which examined the role of expectation in motor strategy modulation in a low load / risk controlled environment. In order to fill this gap in the knowledge base, the MPD testing protocol was used to investigate the role of expectation in motor strategy modulation in a low-load / risk control environment for the first time (Unpublished data, Chapter 5). The key distinction for this study being that the motor behaviour of participants was examined when expecting a familiar, safe and subtle perturbation. Furthermore, the perturbation occurred during SSC activity in a low load / risk environment (sleigh) where the participants’ body weight was offset and the potential for loss of balance or failed loading was minimised.

It is reasonable to hypothesise that under such experimental circumstances participants’ behaviour would not change if expecting a small, familiar safe and subtle change in floor surface height. However, overt changes in behaviour were still observed (Unpublished data, Chapter 5). It is difficult to identify the exact reason for any of the changes under the controlled experimental setting. For example, a statistically significant increase in θ_{vel} was observed bilaterally ($p < .020$) when participants expected a change in floor surface (EH) when compared to baseline hopping (BH). There is a suggestion in the literature that proprioceptive acuity may be velocity dependent as passive movement detection thresholds can be reduced up to tenfold with increased velocity of the imposed movement (Hall and McCloskey 1983; Refshauge, Chan et al. 1995). However, the speeds examined in such studies were very slow. For example, Hall and McCloskey (1983) examined this function over a range of velocities from 10 to 80 °/s at various joints. The θ_{vel} during a SSC far exceeds those explored previously. For example, during BH the mean stretch velocity of the ankle were much higher (Right = 264.1 +/- 71.0 °/s; Left = 276.1 +/- 79.3 °/s) than those tested elsewhere. Furthermore, the stretch velocity increased during EH to 344.2 +/- 48.3 °/s for the right ankle and 337.6 +/- 57.1 °/s for the left ankle (all $p < .020$). It is not clear whether this change was an attempt to cater for the mechanical change that was expected or if increasing stretch velocity represented an effort to become more sensitive as they were

instructed to try to inform the testers if they felt the change in surface height. More research is required to explore the relationship between stretch velocity and proprioceptive acuity during SSC activity.

In the same study an increase in stiffness was observed on the side of the impending change during EH ($p = .013$). Under normal and high load scenarios such feedforward stiffness modulation is considered a protective response to mitigate against the risk of a failed loading scenario e.g. injury, falling over or significant compromise to COM dynamics (Ferris, Liang et al. 1999; McDonagh and Duncan 2002; Moritz and Farley 2004; Hobara, Omuro et al. 2007; Masahiro and Shingo 2010; Yeadon, King et al. 2010; Leukel, Taube et al. 2012). However, stiffness modulation was observed in a low risk environment. Further research should consider modifying the cuing and sides of perturbation of the participants to explore the effect of cuing on stiffness modulation. Once again, it cannot be discounted that the observed changes in stiffness could be an attempt to increase sensitivity to the change in surface height. To investigate this new concept, participants could be requested to perform the test at different stiffness settings and any changes in MPD performance examined. This could be achieved by altering their mass, or contact times as both can directly affect stiffness (Dalleau, Belli et al. 2004; Hobara, Kanosue et al. 2007). Furthermore, it could be investigated whether “stiffer” people are more sensitive during the MPD testing protocol.

Finally, there was a clear pattern of increased EMG activity to the muscles of the ankle during EH in the absence of an increase in the work done. Once again, this is consistent with previous findings of increased activation of multiple leg muscles including Sol, MG and TibAnt prior to contact when hoppers expected an increase in floor stiffness (Moritz and Farley 2004). The critical difference in the research paradigm of this thesis is that the intention was not to maintain centre of mass dynamics in the face of a significant threat or external perturbation. The primary instruction to the participants was to search for the subtle perturbation that they may or not feel. In fact, they were primed to the direction (up or down) and the side of the surface change (right). Therefore, it cannot be discounted that the above differences in motor behaviour observed for EH may be due to the participants' efforts to try to perceive the impending perturbation. Therefore, the observed changes may actually represent a sensory searching strategy rather than the traditionally described protective motor strategies. The author is unaware of any other research which has described the potential for a sensory searching strategy contributing to the behavioural changes observed when expecting an external perturbation.

8.5 What causes perception of change in the foot floor-interface?

This thesis represents the seminal experimental paradigm that considers participants' cognitive perception of changes in floor surface height whilst also examining the motor responses to the perturbation (Unpublished data, Chapters 6 & 7). The author is similarly unaware of any other research which has examined the motor responses to subliminal perturbations. Proprioception is traditionally triggered by movement or changes in position (Proske and Gandevia 2012). Therefore, the derived variables in this thesis were based upon the mechanical configuration of the ankle during the contact phase of hopping - for example stretch amplitude, stretch velocity, stiffness and ankle angle at contact. It was hypothesised from reviewing the proprioception literature that significant changes in these variables may trigger a change in configuration of the limb, and hence perception of the external perturbation. A key finding of this body of work is the absence of statistically significant differences in these kinematic variables when comparing perceived (P; 36mm) and subperceptual (SubP; 6mm) height changes to EH (Unpublished data, Chapter 8). Similarly, comparing EMG from the SubP and P conditions and further comparison of both to EH EMG data revealed no definitive changes that would suggest that muscle activity triggered perception of the 36mm height increase (Unpublished data, Chapters 6 & 7). Therefore, it remains unclear what factor(s) trigger perception during the MPD test as no observable changes in the configuration of the ankle or its motor behaviour could be attributed to successful perception of the external perturbation.

The participants in the present body of work registered 100% perception of the 36mm height change and 100% non-perception of the 6mm height change. However, there was no control for each individual's actual threshold of perception. Performing a similar analysis at each participant's threshold of perception may reveal if there exists a consistent change in any of the above variables that may trigger detection of the change in floor surface height.

8.6 Summary of the thesis

The studies contained in this manuscript aimed to fill some of the gaps in the proprioception literature by merging the experimental constructs commonly used to separately investigate proprioception and dynamic modulation of SSC activity. It is beyond the scope of this single thesis to clarify all aspects of proprioceptive function during the SSC; however this body of work has aimed to establish a robust methodology for examining this broader concept. Firstly, the reliability and stability of human hopping on the custom built sleigh has been examined in detail (Unpublished data Chapter 3; Debenham et al, Submitted Manuscript-

Appendix 1). Using a sleigh hopping model, the Minimal Perceptible Difference test has been established. The theoretical construct of this novel testing procedure has been explored and the reliability of the system demonstrated (Travers, Debenham et al. 2013).

The motor strategies associated with performance of the novel MPD testing protocol have been investigated. The role of expectation of a safe, subtle and familiar perturbation during SSC function in a risk minimised environment has been explored for the first time (Unpublished data, Chapter 5). In addition, the potential drivers of cognitive perception during the MPD test has been investigated thoroughly (Unpublished data, Chapters 6 & 7). Using the MPD paradigm, the findings of these studies have established clear lines for future research in this otherwise unexplored area.

9. Conclusions

There exists a gap in the literature pertaining to proprioception during functional movements of the lower limb. Although multiple studies have investigated the motor strategy responses to external perturbations during SSC function, the concept of participants' cognitive perception of the perturbation has remained unexplored. This thesis has explored the concept of proprioception during dynamic function of the ankle by merging the experimental constructs commonly used to separately investigate proprioception and dynamic modulation of SSC activity. This has been achieved via investigating the perception of change in the foot-floor interface during repeated SSC. To this end, the Minimal Perceptible Difference (MPD) testing protocol has been developed. This body of work has demonstrated the reliability and validity of the sleigh hopping system and the associated derived variables. The theoretical construct of the MPD testing protocol has been explored and justified and the factors influencing participants' threshold in the test have been investigated. The reliability of performance during the MPD test and strategy dependent sensitivity have been described. Finally, the MPD protocol has been used to investigate the motor strategies associated with expectation of and perception of an impending perturbation in a risk minimised environment.

The following are the specific conclusions from each of the chapters, supplemental chapters and analysis contained in the appendices:

In Chapter 3, leg stiffness and the kinematic profile of the ankle were compared during self-paced upright and sleigh hopping. Both are highly reliable tasks for examination of ankle stiffness and kinematic variables on a within and between day basis. A strong correlation between stiffness and stretch amplitude independent of the loading suggesting that individuals have a default setting that they apply to each task. Furthermore, the temporal stability of sleigh hopping has been demonstrated in Appendix 1. Thus, self-paced sleigh hopping represents an appropriate methodology for modelling normal dynamic function whilst negating the confounding features of upright hopping such as fatigue and balance.

Supplemental Chapter 1 further supports the use of a sleigh hopping model as an appropriate methodology for investigating the SSC performance at the ankle by validating the use of the

leg stiffness estimate during sleigh hopping. Furthermore, it has been demonstrated that the leg stiffness estimate represents a measure of ankle stiffness under the present methodology.

Chapter 4 presents the MPD test as a novel test for assessing detection of floor height changes during sleigh hopping. This represents a change towards investigating the concept of proprioception in more functional tasks by using a repeated SSC model. Furthermore, the MPD test has been demonstrated as reliable over time and is therefore an acceptable tool for assessing this function within and across test occasions. Greater sensitivity of the MPD test during bilateral hopping may reflect finely tuned sensory requirements for the maintenance of upright stance which may be much less relevant during gait.

The use of a sliding floor mechanism causes methodological challenges with respect to identifying event markers during hopping due to force plate artefact. The use of kinematic data to measure temporal events in hopping represents a novel solution and is explored in Supplementary Chapter 3. Comparison of kinematic data to define event markers to the gold standard (force plates) demonstrated that the Vicon 3-D Motion Analysis System is highly accurate and reliable for measuring flight and contact times during sleigh hopping. A small and systematic bias in the duration of these hopping events may be accounted for by force-plate threshold cut-off.

Supplemental Chapter 2 investigated modulation of consciously controlled increase in ankle stiffness during sleigh hopping. Leg stiffness was increased by a temporal compression of the motor pattern of the ankle plantarflexors in order to drive a stiffer hopping performance. Importantly, increased co-activation of the ankle muscles was not observed and was not a strategy for stiffness modulation under conscious drive. These findings contribute to lower limb stiffness modulation theories and are also consistent with the observed motor control strategies in some spinal pain syndromes. Therefore, there appears to be at least two fundamental motor control strategies to modulate joint stiffness and that (as observed on spinal pain models) in the absence of co-activation strategies a preferred control mechanism is temporal modulation of the feed forward responses.

Chapter 5 has demonstrated that the expectation of a change in the foot – floor interface produced overt changes in participants' baseline bilateral hopping strategy. These changes

included reduced contact time and increased stretch velocity bilaterally, and increased stiffness on the side of the impending perturbation. EMG signal amplitude increased in both agonist and antagonist muscles of both ankles in response to the expectation. Different activation patterns observed for the mono-articular and bi-articular muscles may represent a sensory searching strategy rather than a motor response to cater for the impending perturbation. Future studies utilising external perturbations during repeated stretch shortening cycles should account for the role of expectation alone on baseline hopping strategies.

The combined findings of Chapters 6 and 7 demonstrated no significant differences in range or position related outputs for the ankle when either subliminal or consciously perceived perturbations were introduced. These findings question the utility of traditional testing methods of measuring proprioception in the context of dynamic function of the ankle. Further research is required to determine if the MPD test measures the same domain as traditional proprioceptive tests. It is clear that future research is required to determine what factors drive perception of changes in floor height during the MPD test.

10. Supplementary Chapter 1: Generalised model for estimates of leg stiffness during low-load plyometrics

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Contributing authors (based on the ICMJE criteria)

Tiffany L. Grisbrook¹, Lei Cui², Mervyn J. Travers¹, James Debenham^{1, 3} & Garry T. Allison¹

Affiliations:

¹School of Physiotherapy and Exercise Science, Curtin University

²Department of Mechanical Engineering, Curtin University

³ School of Physiotherapy, University of Notre Dame

Abstract

Hopping is frequently used to assess leg stiffness; although upright hopping is not always practical in populations where lower-limb function is impaired. Numerous studies have assessed leg stiffness on sledge jumping systems (SJS); however these SJS often don't represent upright hopping. We have developed a SJS that allows individuals to mimic hopping in a low-load manner. A field based measurement of leg stiffness, that uses contact and flight time modelling has been validated in upright hopping, however this method has not been validated to allow for changes in gravity associated with hopping on an inclined SJS. This study provides the mathematical proof that the formulae can be generalised in the context of SJS stiffness estimates. 3D motion analysis and force plate signals were used to validate estimates of lower limb stiffness across a wide range of hopping frequencies. Further this study determined which individual joint spring was the main correlate to limb stiffness.

Results demonstrated that the contact and flight time method of estimating stiffness was strongly correlated to the traditional spring-mass modelling of stiffness, during hopping on the SJS ($r^2=0.95$). Further we found that ankle stiffness is the major determinant of leg stiffness during low load sledge hopping ($r^2= 0.85$). These results suggest that lower limb stiffness can be determined on an inclined SJS, and the field method of estimating stiffness may be a useful tool in determining rehabilitation strategies in clinical environments where rehabilitation incorporates stiffness modulation and in individuals where only low-load plyometric activities are possible.

Introduction

The lower-limb is often referred to as having spring-like characteristics during several locomotive tasks including running, hopping and jumping (Farley and Ferris, 1998; Kuitunen et al., 2010). Therefore the musculoskeletal elements of the lower-limb can be modelled as a spring-mass model, consisting of a body mass and a single linear leg spring (Blickhan, 1989). The ability of the spring-system to resist any change in length when force is applied to it is referred to as stiffness (Kuitunen et al., 2010). Hopping is frequently used to assess leg stiffness (Blickhan, 1989; Farley et al., 1991; Farley and Morgenroth, 1999; Dalleau et al., 2004; Kuitunen et al., 2010; Hobara et al., 2011). Overall leg stiffness is dependent on a combination of the stiffness of each of the individual joint springs. It has been reported that leg stiffness is primarily influenced by ankle stiffness during upright submaximal hopping (Farley and Ferris, 1998; Farley and Morgenroth, 1999; Yen and Chang, 2010; Kim et al., 2013). In contrast, Hobara et al. (2009), found that knee stiffness was the major determinant of leg stiffness in maximal height hopping (Hobara et al., 2009). Kuitunen et al. (2010) concluded that in hopping, the joints play different roles. They documented that leg stiffness modulation is sensitive to changes in ankle stiffness, whereas knee stiffness plays an important role in regulating hopping intensity and height (Kuitunen et al., 2010). More recently, it has been reported that the major determinant of leg stiffness is altered according to hopping frequency, where the knee explains more of the variance at 1.5 Hz, and the ankle is the major influence of leg stiffness when hopping at 2.2 Hz or above (Hobara et al., 2011). The literature is dominated by hopping with load much greater than body mass. Therefore little is known about the contribution of each of the individual joint springs during low-load plyometric activities.

Although upright hopping is conceptually relevant to assess changes in leg stiffness, it is not always practical in populations where lower-limb function is impaired. Numerous studies have assessed the stretch-shortening cycle during hopping utilising various sledge jumping systems (SJS). Such SJS are advantageous over upright hopping, as they allow for reliable and controlled measurement of mechanical efficiency and motor correlates in a dynamic environment (Kramer et al., 2010; Furlong and Harrison, 2013). The majority of these studies have utilised a SJS whereby participants are seated in an upright position, and are able to slide along rails that are inclined approximately 20° from the horizontal (Aura and Komi, 1986; Kyrolainen and Komi, 1995; Flanagan and Harrison, 2007; Ertelt and Blickhan, 2009). There have been efforts to improve SJS to better represent natural upright hopping and jumping (Bubeck and Gollhoffer, 2001; Kramer et al., 2010; Furlong and Harrison, 2013). While these SJS provide reliable measures of leg stiffness, they only

provide similar movement patterns to upright hopping, and don't require subjects to rely on their own motor control. Therefore our research group developed a novel SJS to allow individuals to mimic upright hopping in a low-load well controlled manner (Gibson et al., 2013; Travers et al., 2013). Individuals lie on the low-friction sledge, which glides along rails inclined 20° from the horizontal. In this SJS, individuals are able to perform natural hopping and jumping, with a reduced body mass. Such a device would allow individuals with loss of muscle capacity to maintain a low amplitude hopping strategy of submaximal loading for a period of time. This may provide a good framework to test repetitive motor control strategies during low-load plyometric activities, which could be utilised in rehabilitation.

Previous studies have utilised various SJS's to examine both leg and joint stiffness. In all of these studies stiffness has been calculated in a laboratory using the spring-mass model which requires the use of force platforms and high-speed kinematic analysis systems. However if the SJS is going to be transferable into clinical or field environments a simplified method of calculating stiffness is required. Recently, Dalleau et al. (2004) proposed a field based measurement of leg stiffness where stiffness is calculated using body mass, ground contact time and flight time which are measured from a contact mat. Dalleau et al. (2004), validated this method against the reference model, finding the two methods of calculating leg stiffness were significantly correlated in maximal ($r=0.98$) and submaximal ($r=0.94$) upright hopping in adults. Lloyd et al. (2009), confirmed that Dalleau's method was significantly correlated to the reference method in submaximal ($r=0.92-0.95$) upright hopping in youths, although the two methods were not related during maximal hopping in this group ($r=0.59$). The Dalleau model has not been generalised to allow for the change in gravity associated with hopping on a SJS. Therefore the purpose of this study was to report the generalised formulae of calculating lower limb stiffness from flight and contact time modelling. This could have significant implications for the utility of various SJS in clinical or field environments (See Chapters 1, 3 and 11). We specifically tested the hypothesis that the ankle represents the primary determinant of leg stiffness when hopping on the SJS. Furthermore, we tested the hypothesis that increased hopping frequency correlates with increased leg stiffness. Finally, we tested the hypothesis that Dalleau's estimate of leg stiffness correlates with ankle stiffness when hopping on the SJS.

Methods

Participants

Five healthy adults, with no functional limitations or history of surgery in the lower extremities, volunteered for this study. Participants included two males and three females (age: 27.56 (\pm 1.32) years, body weight: 73.68 (\pm 21.58) kg, height: 177.6 (\pm 14.42) cm). This study was approved by the institutions Human Research Ethics Committee and all participants provided written informed consent prior to participation.

Instrumentation

Participants hopped on the low-friction sleigh, which slid along rails reclined to 20°, and the weight of the sled was offset so that each participant could perform repetitive plyometric activities, at a reduced load (Gibson et al., 2013). A 14 camera Vicon MX ® (Vicon Motion Systems, Lake Forest, CA) motion analysis system digitally sampling at 250 Hz and a force plate (AMTI®, Watertown, MA) mounted in the base of the SJS sampling at 1000 Hz, were utilised to analyse hopping kinematics and kinetics. In accordance with a customised lower-limb model, individual or clusters of 3-4 retro-reflective markers, each 16mm in diameter, were affixed to specific anatomical landmarks on the participants with double-sided tape as per Chapter 5.

Protocol

Prior to data collection, demographic and anthropometric variables (age, leg dominance, height, mass and foot length) were recorded. Participants were required to perform a series of unilateral barefoot hops while lying on the SJS; both lower-limbs were assessed in a randomised order. Participants were required to perform familiarisation trials until satisfied they could perform the task correctly.

Three different hopping frequencies were then collected for analysis. Initially the participants were instructed to hop at a self-selected hopping frequency. They were instructed to hop keeping their knee and hip as straight as possible, at a comfortable frequency that they felt they could maintain indefinitely. Once a stable hopping frequency was established, the frequency of hopping was recorded and then input into a digital metronome. The participants then completed five, 10 second, trials of self-selected hopping with an audio cue. The participants were then required to match the metronome frequency, which was set either 25% faster, or 15% slower than the self-selected hopping frequency. This was to ensure the full range of normal hopping frequency was tested under this

validation protocol. The participants completed five, 10 second, trials of hopping at each frequency, for each lower-limb, which were performed in a random order. One minute of rest was required between each hopping frequency, and 30 seconds of rest was allocated between each trial.

Data reduction

Three dimensional data were processed using standard biomechanical procedures, in Vicon Nexus software (Vicon, Oxford Metrics, Oxford, UK). All marker trajectories and ground reaction forces (GRF) were filtered using a Butterworth low pass filter (2nd order, zero lag, initial steady state assumed) at an optimal frequency determined using a residual analysis algorithm (Labview®, National Instruments, Austin Texas) – cut off frequency 14 Hz. Three dimensional motion data was modelled using Vicon BodyBuilder code (Vicon Motion Systems, Lake Forest, CA). The model for calculating joint kinematics and kinetics was adapted from the customised UWA full body model (Reid et al., 2008) as per Chapter 5. The segments modelled were the foot, shank, thigh and pelvis. Joint angles were defined according to the ISB recommendations (Wu and Cavagnagh, 1995). Joint moments and powers were calculated using inverse dynamics with inertial parameters as described by de Lava (de Lava, 1996).

Stiffness calculations

Modelled data was then exported from Nexus for further analysis using a customised program in Labview version 2011 (National Instruments, Austin, TX). This program calculated leg stiffness for each individual hop during each 10 second hopping trial, for all hopping frequencies. Estimates of leg stiffness were calculated using the spring-mass model utilising two different methodologies. The reference method developed by Cavagna et al (1988) was compared to the new generalised method. For Cavagna's method, leg stiffness (K_C) was calculated by dividing the peak vertical GRF (F_{peak}) (adjusted to align with the axis of the SJS) by the maximum displacement of the COM (ΔZ), in alignment with the axis of the SJS, during the ground contact phase (Cavagna et al., 1988).

$$K_C = F_{peak} / \Delta Z$$

Appendix A provides the mathematical proof that the estimates of leg stiffness within the model assumptions of Dalleau et al (2004) is unaffected by the inclination of the SJS. The model uses time in contact (T_c), flight time (T_f), and body mass (M).

$$K_D = \frac{BM \times \pi (T_f + T_c)}{T_c^2 \left[\frac{(T_f + T_c)}{\pi} - \frac{T_c}{4} \right]}$$

Joint stiffness (K_{joint}) of the ankle and knee joints, were calculated using the methods described by Farley and Morgenroth (1999). Where K_{joint} is calculated as the ratio of the peak joint moment (M_{joint}) to the maximum joint angular displacement ($\Delta\theta_{joint}$) in the sagittal plane, during the ground contact phase.

$$K_{joint} = \frac{M_{joint}}{\Delta\theta_{joint}}$$

The combination of the stiffness of the ankle and knee joint springs in series ($K_{ankle+knee}$) was also calculated (Hoque et al., 2006).

$$K_{ankle+knee} = \frac{K_{ankle} K_{knee}}{K_{ankle} + K_{knee}}$$

Statistical Analysis

The mean stiffness of the median six hops for each trial was calculated and used for data analysis. A two way repeated measures ANOVA was used to compare the effects of limb dominance (dominant and non-dominant) and hopping frequency (self-selected, fast and slow) on estimates of leg stiffness calculated using two methodologies (K_C and K_D). Pairwise comparisons were calculated for significant main effects, using the Bonferroni adjustment. Linear regression analyses were conducted to determine the coefficients of determination (r^2) between; K_C and K_D , K_D and K_{joint} , and K_D and $K_{ankle+knee}$ for each subject and for the grouped results. Data was normalised for each individual's self-selected hopping frequency for both stiffness assessments to determine if they were concordant on changes in stiffness associated with hopping frequency changes. Sensitivity of changes (bias and typical error) between K_C and K_D were determined by calculating the mean (95% CI) difference (D)

and the standard deviation of the difference (SD_D), and the typical error ($SD_D / \sqrt{2}$). In all analyses, the alpha level was set to $p < 0.05$.

Results

The effect of limb dominance on leg stiffness

There was no difference in leg stiffness between the dominant and non-dominant legs when calculated using either K_C ($p= 0.281$) or K_D ($p=0.430$) method (Table 1). Therefore results from both legs were analysed simultaneously for the remainder of the statistical analysis.

Table 1: Lower limb stiffness for the dominant and non-dominant leg at three hopping frequencies calculated using the Cavagna (K_C) and Dalleau (K_D) methods

Leg	Hopping Condition	Hopping (bpm)	K_C (kN.m⁻¹)	K_D (kN.m⁻¹)
Dominant (n=17)	S-Selected	76.31 (± 11.78)	7.78 (± 3.41)	7.49 (± 4.51)
	Fast	90.75 (± 14.01)	10.47 (± 3.05)	10.45 (± 4.76)
	Slow	68.77 (± 10.62)	7.97 (± 2.90)	7.45 (± 4.04)
Non-dominant (n=15)	S-Selected	77.04 (11.15)	8.95 (± 3.72)	8.73 (± 4.61)
	Fast	91.77 (13.59)	11.90 (± 3.55)	12.01 (± 6.44)
	Slow	69.43 (10.05)	9.06 (± 3.21)	8.57 (± 4.05)

S-Selected = Self selected or preferred hopping frequency

The effect of hopping frequency on leg stiffness

The participants self-selected hopping frequency was 76.31 (± 11.78) bpm (range= 65.68- 95.9) for the dominant leg and 77.04 (± 11.15) bpm (range= 68.25- 95.54) for the non-dominant leg (Table 1). The total range of hopping frequency observed across all three hopping frequencies was 59.19- 114.05 bpm. This range reflects a large proportion of the normal range (52 to 134) previously observed on the same sleigh.

Hopping frequency significantly affected both K_C ($F(2, 60) = 58.884, p < 0.001$) and K_D ($F(2, 60) = 42.846, p < 0.001$). Post hoc tests revealed that leg stiffness significantly increased from self-selected hopping frequency to fast (125%) hopping frequency for both K_C (mean difference $2.82 \text{ kN}\cdot\text{m}^{-1}$, $p < 0.01$) and K_D (mean difference $3.122 \text{ kN}\cdot\text{m}^{-1}$, $p < 0.001$). Likewise, both K_C (mean difference $2.668 \text{ kN}\cdot\text{m}^{-1}$, $p < 0.001$) and K_D (mean difference $3.122 \text{ kN}\cdot\text{m}^{-1}$, $p < 0.001$) significantly increased from the slow hopping frequency compared with the fast hopping frequency. No systematic differences were detected for self-selected and slow (85%) hopping.

Relationship between K_C and K_D

Figure 2 displays an example of leg stiffness values for each individual hop for one participant (Subject 3) for all trials, at all frequencies. For Subject 3, K_C and K_D are significantly correlated ($r^2 = 0.94$). Raw data comparisons for all subjects resulted in an estimate of typical error ($SD_{\text{Diff}} / \sqrt{2}$) 1.30 (95% CI 1.15 to 1.49), a mean difference of 0.31 (95% -0.03 to 0.64) and an ICC 0.894 (95% CI 0.851 to 0.925).

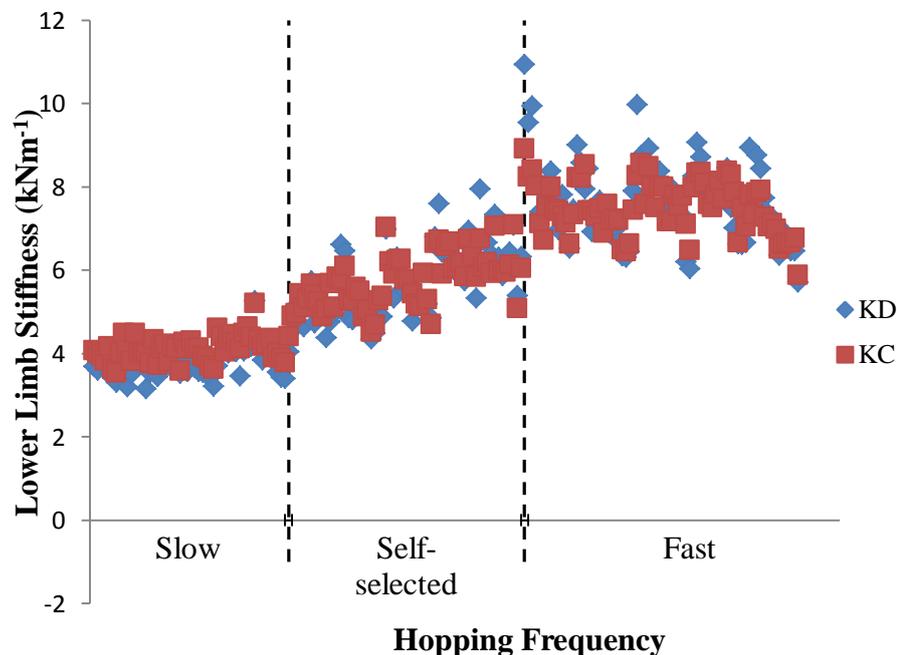


Figure 2: Lower limb stiffness calculated using Cavagna (K_C) and Dalleau (K_D) methods, for each individual hop for one participant at all hopping frequencies

It is evident that stiffness is influenced by hopping frequency, which has been statistically confirmed above. Therefore, to compare coefficients of determination between subjects, stiffness values were normalised to the mean stiffness (K_C or K_D) for the self-selected hopping frequency for both limbs.

Normalised stiffness values for K_C were very strongly correlated with K_D for all subjects (mean $r^2 = 0.96$, range = 0.94- 0.98) (Figure 3a). When all subjects normalised stiffness data were combined $r^2=0.95$ (Figure 3b). The mean difference between tests was 0.69% (CI -1.95 to 0.56%), there was a typical error of 4.81% (CI 4.16 to 5.52%) corresponding to an ICC of 0.977 (CI 0.964 to 0.983).

Relationship between K_D and K_{joint}

Normalised K_D and normalised K_{ankle} were moderately to very strongly correlated for all subjects (mean $r^2 = 0.82$, range: 0.63-0.95) (Figure 4a). When all subjects data were combined $r^2 = 0.85$ (Figure 4b). There was no relationship between K_D and K_{knee} (mean $r^2=0.19$) or between K_D and $K_{ankle+knee}$ (mean $r^2 = 0.11$).

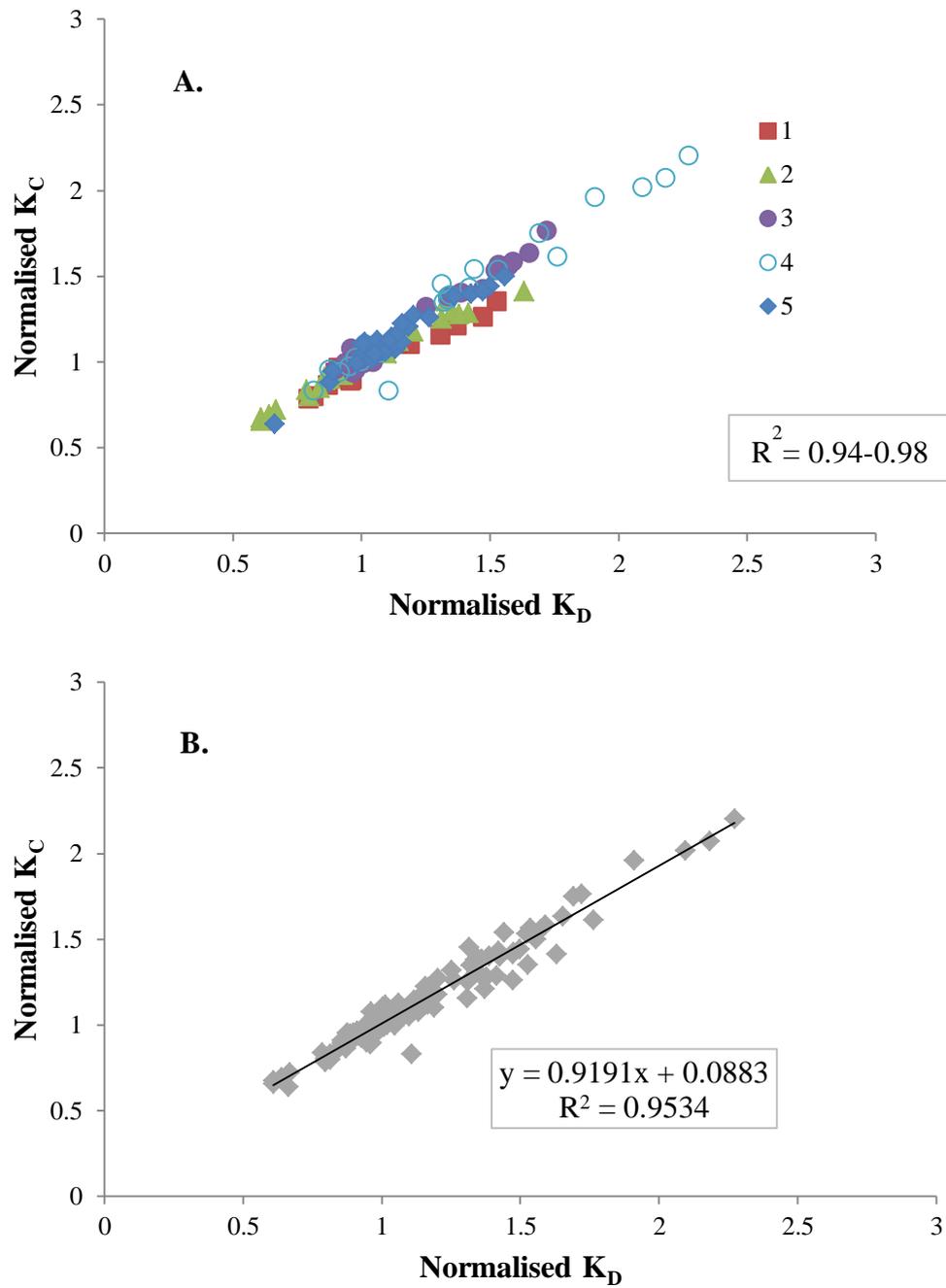


Figure 3: The correlation between normalised lower limb stiffness calculated using the Cavagana (K_C) and Dalleau (K_D) methods for: (A) each individual subject and (B) all subjects combined

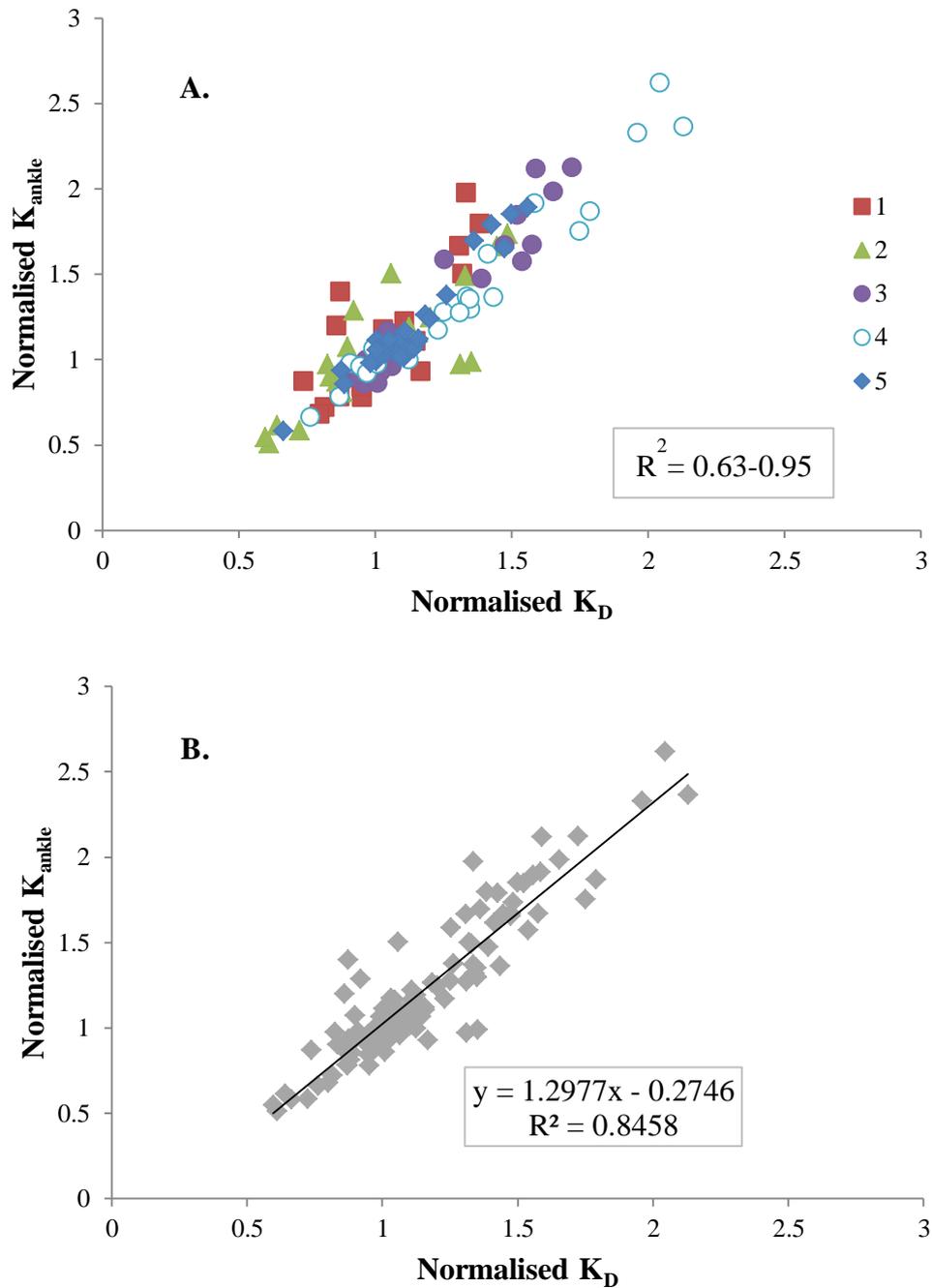


Figure 4: The correlation between ankle joint stiffness (K_{ankle}) and normalised lower limb stiffness calculated using the Dalleau method (KD) for: (A) each individual subject and (B) all subjects combined.

Discussion

This study has provided a mathematical validation of the Dalleau et al. (2004) model for estimates of leg stiffness while being assessed on an inclined sleigh system. This provides some evidence that within the assumptions of the original model the equation can be used in

a sleigh system. Future research however will need to examine the generalizability of the model that would account for the sleigh inertial and friction characteristics as well as higher load plyometrics.

It is generally accepted that leg stiffness increases with increasing hopping frequency (Farley et al., 1991; Ferris and Farley, 1997; Dalleau et al., 2004; Hobara et al., 2011), which was true for the current study in an inclined task. Both K_C and K_D significantly increased as hopping frequency increased from the self-selected to the fast hopping frequency. This supports the findings of Dalleau et al. (2004) that both their surrogate and Cavagna's methods of estimating leg stiffness increased with increasing upright hopping frequency. They concluded that the surrogate method could therefore be used to examine individual variations in leg stiffness induced by changes in hopping frequency, without the need for force measurement. This current study extended this surrogate method to activities on a sleigh system. The literature demonstrates that leg stiffness plays an important role in optimising the efficiency of locomotion. Farley et al. (1991) found that during upright hopping at a preferred frequency, ground contact time was longer and presumably muscle efficiency was optimised when compared to faster frequencies. This is further supported by previous literature that has demonstrated increased leg stiffness to be associated with faster running velocities (Arampatzis et al., 1999), and inversely related to the metabolic cost of running (Dalleau et al., 1998). However, using a sleigh system the capacity of the individual participant to modulate the hopping frequency (lower limb stiffness) may be influenced by motor control aspects as opposed to optimising the muscle metabolic capacity. Indeed on a sleigh with an inclination of 20 degrees individuals can hop at the equivalent of 34% of their body mass. This may improve the degrees of freedom for people in rehabilitation with limited muscle capacity or co-ordination by allowing them to experience the stretch shortening cycle and perform low-load plyometric rehabilitation activities. This may reflect opportunities for neurological rehabilitation that otherwise would be limited by an inability to hop vertically.

The concordance between the criteria (K_C) and surrogate model of normalised data based on self-selected pace demonstrates that within subjects changes in stiffness can be detected with great sensitivity with typical errors of less than 5%. Further research is needed into measuring the metabolic cost of hopping at different frequencies on a SJS, and to determine the validity of the model when higher load plyometrics are performed in an unloaded scenario.

The results of this study support those of Flanagan and Harrison (2007) and Bachman et al., (1999) who detected no side to side lower limb stiffness asymmetry in sleigh rebounding and running respectively.

While it has been reported that leg stiffness is primarily influenced by ankle stiffness during upright submaximal hopping (Farley and Ferris, 1998; Farley and Morgenroth, 1999; Yen and Chang, 2010; Kim et al., 2013), little was known about the contribution of each of the individual joint springs during low-load plyometric activities performed on a SJS. Further it was unknown if joint stiffness was related to the generalised contact and flight time modelling of leg stiffness. The results of the current study demonstrate that K_{ankle} is well correlated to K_D whereas there is no relationship between K_{knee} and K_D . This demonstrates that K_{ankle} is the major determinant of K_D during hopping in the novel SJS, which suggests that individuals use a similar strategy to modify leg stiffness in low-load hopping that they do in upright submaximal hopping. However, it must be noted that the task of hopping on the sleigh (unloaded body mass) means that much less force is required to achieve displacement of the COM and therefore foot clearance. A future direction of research would be to examine the relative contributions of the lower limb joints if the vertical height change was matched between upright hopping and sleigh hopping. This direction of research is necessary as Kuitunen et al (2010) found that increasing hopping intensity (increasing vertical GRF) during bilateral upright hopping, resulted in K_{ankle} remaining unchanged, while K_{knee} increased significantly. It is possible that the power production across the knee and hip contributing to the high power production in the lower limb kinetic chain may necessitate a different modulation strategy of K_{ankle} . This hypothesis is further supported by one individual in this data set. Subject 1 hopped with increased effort in comparison to the other subjects, and as such required increased contribution from the knee joint. This is further supported by Hobara et al (2009), who found that the knee joint was the major determinant of stiffness during maximal effort hopping. Therefore the ankle joint appears to play an important role in stiffness adjustment during submaximal hopping, while the knee joint plays a more important role in maximal hopping, in both upright and SJS hopping. Nevertheless, in this study we see that even when the two joint springs in series were combined ($K_{\text{ankle+knee}}$), ankle stiffness was still a stronger predictor of leg stiffness. Modulation of stiffness around the ankle during submaximal hopping can therefore be determined by contact and flight time modelling. This provides further support for the use of this SJS in clinical or field environments. Identifying major determinants of leg stiffness during hopping is helpful in identifying the most effective training methods (Hobara et al., 2011). This is particularly relevant for low-load hopping, given that individuals with injuries or neurological disorders

may not be able to hop upright, but would be able to maintain a low amplitude hopping strategy of submaximal loading for a period of time on the SJS. This may provide a good framework to determine optimal rehabilitation strategies using low-load plyometric activities where high numbers of repetitions may be the rehabilitation goal for motor retraining as opposed to power development training.

A limitation of this study is that we undertook a validation study model where a limited number of participants were instructed to change their normal hopping frequencies that covered a large proportion of the previously observed self-selected hopping range. This range is based on self-selected ranges and may not be generalizable to high load power training. The majority of the literature in this field addresses upright hopping, submaximal and maximal jumping (including drop jumps) and therefore comparisons with the previous literature are difficult to make. The advantage of the sleigh is that the balance characteristics (vestibular input of sign leg hopping) are reduced however the inertial and frictional factors contributing to the individual's performance have not been modelled and therefore warrants further investigation.

Conflict of interest statement

All authors concur that there are no financial or personal relationships with other people or organisations that may inappropriately influence the content of the work submitted for publication.

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Mathematical Proof

Background

When an individual hops on an unloaded sleigh the impact of the inclination relative to gravity alters the behaviour of the individual. The flight times are greater and the load is less. There are concerns that the Dalleau et al (2004) model that has been validated against the vertical may not be valid for the sleigh. One criticism is that the flight times are altered due to the decrease in gravity. This is true however the follow proof shows that within the assumptions made by Dalleau in the original derivation the impact of the reduced gravitational force has no change to the original formulae.

Leg Stiffness Calculations on a Sleigh

θ is the angle of inclination from the horizontal.

Step 1: Assume the force-time signal is a sine wave

$$F(t) = F_{\max} \sin\left(\frac{\pi}{T_c} t\right) \quad (1)$$

where F_{\max} is the peak force, T_c is the contact time and is the half period of the sine wave.

Step 2: Determine the maximal force F_{\max}

The momentum change during the contact time is

$$\int_0^{T_c} (F(u) - Mg \sin \theta) du = M \Delta v = Mg \sin \theta T_f \quad (2)$$

where M the body mass, g gravitational acceleration, θ the sleigh angle, T_c the contact time, T_f the flight mean time and T_f is calculated by the mean of flight time before and after one contact.

Substituting Eq. (1) into Eq. (2) yields

$$\begin{aligned} \int_0^{T_c} \left(F_{\max} \times \sin\left(\frac{\pi}{T_c} u\right) - Mg \sin \theta \right) du &= Mg \sin \theta T_f \\ \Rightarrow -F_{\max} \frac{T_c}{\pi} \cos\left(\frac{\pi}{T_c} u\right) \Big|_0^{T_c} - Mg \sin \theta T_c &= Mg \sin \theta T_f \\ \Rightarrow -F_{\max} \frac{T_c}{\pi} (-1 - 1) &= Mg (T_f \sin \theta + T_c \sin \theta) \\ \Rightarrow F_{\max} &= \frac{Mg\pi}{2} \sin \theta \left(1 + \frac{T_f}{T_c} \right) \end{aligned} \quad (3)$$

Step 3: Calculate the acceleration

The Newton's second law gives the acceleration as:

$$Ma(t) = \sum F = F(t) - Mg \sin \theta = F_{\max} \sin\left(\frac{\pi}{T_c} t\right) - Mg \sin \theta \quad (4)$$

Thus the acceleration can be obtained as

$$a(t) = \frac{F_{\max}}{M} \sin\left(\frac{\pi}{T_c} t\right) - g \sin \theta \quad (5)$$

Step 4: Calculate the velocity

Integrating Eq. (5) gives the velocity $v(t)$ as

$$v(t) = \int_0^t a(u) du + v(0) = \int_0^t \left(\frac{F_{\max}}{M} \sin\left(\frac{\pi}{T_c} u\right) - g \sin \theta \right) du + v(0) \quad (6)$$

where $v(0)$ is the downward vertical velocity at the moment of contact. The velocity can be obtained by integrating the above equation:

$$\begin{aligned} v(t) &= -\frac{F_{\max}}{M} \frac{T_c}{\pi} \cos\left(\frac{\pi}{T_c} u\right) \Big|_0^t - g \sin \theta \Big|_0^t + v(0) \\ &= -\frac{F_{\max}}{M} \frac{T_c}{\pi} \cos\left(\frac{\pi}{T_c} t\right) + \frac{F_{\max}}{M} \frac{T_c}{\pi} - g \sin \theta t + v(0) \end{aligned} \quad (7)$$

Knowing that the velocity is zero at the middle of the contact yields

$$v\left(\frac{T_c}{2}\right) = 0 = -\frac{F_{\max}}{M} \frac{T_c}{\pi} \cos\left(\frac{\pi}{T_c} \frac{T_c}{2}\right) + \frac{F_{\max}}{M} \frac{T_c}{\pi} - g \sin \theta \frac{T_c}{2} + v(0) \quad (8)$$

Solving for $v(0)$ from the above equation gives

$$v(0) = g \sin \theta \frac{T_c}{2} - \frac{F_{\max}}{M} \frac{T_c}{\pi} \quad (9)$$

Substituting Eq. (9) into Eq. (7) yields the velocity

$$\begin{aligned} v(t) &= -\frac{F_{\max}}{M} \frac{T_c}{\pi} \cos\left(\frac{\pi}{T_c} t\right) + \frac{F_{\max}}{M} \frac{T_c}{\pi} - g \sin \theta t + v(0) \\ &= -\frac{F_{\max}}{M} \frac{T_c}{\pi} \cos\left(\frac{\pi}{T_c} t\right) + \frac{F_{\max}}{M} \frac{T_c}{\pi} - g \sin \theta t + g \sin \theta \frac{T_c}{2} - \frac{F_{\max}}{M} \frac{T_c}{\pi} \\ &= -\frac{F_{\max}}{M} \frac{T_c}{\pi} \cos\left(\frac{\pi}{T_c} t\right) + g \sin \theta \left(\frac{T_c}{2} - t\right) \end{aligned} \quad (10)$$

Step 5: Calculate the vertical displacement

Assuming $z(0) = 0$ and integrating Eq. (10) yield the vertical displacement:

$$\begin{aligned}
 z(t) &= \int_0^t \left(-\frac{F_{\max}}{M} \frac{T_c}{\pi} \cos\left(\frac{\pi}{T_c} u\right) + g \sin \theta \left(\frac{T_c}{2} - u\right) \right) du + z(0) \\
 &= -\frac{F_{\max}}{M} \left(\frac{T_c}{\pi}\right)^2 \sin\left(\frac{\pi}{T_c} u\right) \Big|_0^t + \frac{g}{2} \sin \theta (T_c t - t^2) + 0 \\
 &= -\frac{F_{\max}}{M} \left(\frac{T_c}{\pi}\right)^2 \sin\left(\frac{\pi}{T_c} t\right) + \frac{g}{2} \sin \theta (T_c t - t^2)
 \end{aligned} \tag{11}$$

In order to calculate the stiffness, the total displacement at the middle of the contact is calculated at $t = T_c/2$:

$$z\left(\frac{T_c}{2}\right) = -\frac{F_{\max}}{M} \left(\frac{T_c}{\pi}\right)^2 + \frac{g}{8} \sin \theta T_c^2 \tag{12}$$

Step 6: Calculate the stiffness

The stiffness is the ratio of the peak force to the total displacement:

$$K = \frac{F_{\max}}{z(0) - z\left(\frac{T_c}{2}\right)} \tag{13}$$

Substituting the peak force in Eq. (3) and the total displacement in Eq. (12) into the above equation yields

$$\begin{aligned}
 K &= \frac{F_{\max}}{\frac{F_{\max}}{M} \left(\frac{T_c}{\pi}\right)^2 - \frac{g}{8} \sin \theta T_c^2} = \frac{F_{\max} (8M\pi^2)}{T_c^2 (8F_{\max} - gM\pi^2 \sin \theta)} \\
 &= \frac{\frac{Mg\pi}{2} \sin \theta \left(1 + \frac{T_f}{T_c}\right) (8M\pi^2)}{T_c^2 \left(8Mg \frac{\pi}{2} \sin \theta \left(1 + \frac{T_f}{T_c}\right) - gM\pi^2 \sin \theta\right)} \\
 &= \frac{M\pi(T_c + T_f)}{T_c^2 \left(\frac{T_c + T_f}{\pi} - \frac{T_c}{4}\right)}
 \end{aligned} \tag{14}$$

11. Supplementary Chapter 2: Consciously controlled leg stiffness modulation is governed by feed-forward responses

Introduction

The stiffness of the lower limb in human locomotion is commonly modelled using a spring-mass representation (Blickhan 1989). The ‘stiffness’ of the ‘leg-spring’ refers to the resistance of the limb (incorporating all joints) to a change in length (Brughelli and Cronin 2008). This is achieved by the complex interaction between the passive articular and periarticular tissues of each lower limb joint and the level of muscle activation modulated for any given task (Yen, Auyang et al. 2009; Yeadon, King et al. 2010) via the peripheral and central nervous systems (Santello 2005). The concept of lower limb stiffness modulation forms a basis of our understanding of human gait (Farley and González 1996; Butler, Crowell et al. 2003; Hof 2003). Importantly, during gait, lower limb stiffness is not constant and the ability to modulate stiffness in order to adapt to the demands placed upon the limb by the non-homogenous external environment is a common experimental paradigm (McMahon and Cheng 1990; Ferris and Farley 1997; Ferris, Liang et al. 1999; Moritz and Farley 2004; Moritz, Greene et al. 2004; Müller and Blickhan 2010).

Leg stiffness modulation has been demonstrated in experimental settings using a variety of external challenges including hopping and running on surfaces of varying rigidity (Ferris and Farley 1997; Ferris, Liang et al. 1999; Moritz, Greene et al. 2004), running on uneven surfaces (Grimmer and Blickhan 2006; Müller and Blickhan 2010), reduced visual input during hopping (Hobara, Omuro et al. 2007) and even reduced cutaneous feedback (Folkowski, Bishop et al. 2005). All of these tasks are characterised by repeated stretch-shortening cycles (SSC) (Komi 2000; Ishikawa and Komi 2007) and focus on the capacity of humans to adjust leg stiffness to accommodate for different environmental challenges for safe and efficient transition of the centre of mass (Farley and Ferris 1998). This modulation can be achieved via increasing muscle activity in preparation for the subsequent foot strike or landing perturbation (Ferris, Liang et al. 1999). Thus, repeated hopping under different loading rates and intensities forms a model to investigate stiffness during the SSC (Farley, Blickhan et al. 1987; Blickhan 1989; Ferris and Farley 1997; Farley and Morgenroth 1999; Dalleau, Belli et al. 2004; Allison, Utsunoniya et al. 2005; Bobbert and Richard 2011).

The ankle joint is considered the major determinant of total lower limb stiffness during low load tasks such as hopping and walking (Farley and Morgenroth 1999; Moritz, Greene et al. 2004). A modulation strategy of co-activation of agonist and antagonist muscles is a well-established mechanism to increase stiffness (Blickhan 1989) and has been investigated in the lower limb in landing studies (Santello 2005; Yeadon, King et al. 2010), at the knee following anterior cruciate ligament repair (Bryant, Newton et al. 2009) and in the lumbar region in the presence of pain and pain free states (van Dieën, Selen et al. 2003; Moseley, Nicholas et al. 2004; Hodges, van den Hoorn et al. 2009; Morris, Lay et al. In Press: 2013). In the case of the ankle joint the agonist – antagonistic synergy is formed by the Triceps Surae and the Tibialis Anterior muscles (Hobara, Kanosue et al. 2007). It has been hypothesised that co-activation strategies represent an attempt to increase stiffness in order to reduce the risk associated with control decisions made in an environment of reduced sensory input to inform motor performance (Stroeve 1996). For example, increased co-activation strategies at the ankle have been observed during landing amongst basketball players with compromised proprioceptive ability (Fu and Hui-Chan 2007). This may imply that a co-activation strategy to increase “stiffness” is a safer option than the unknown outcome of negotiating a challenge with compromised sensory feedback.

To date, the literature pertaining to lower limb stiffness modulation via co-activation uses unexpected and external challenges as a model to observe changes in limb / joint stiffness. In contrast however, conscious effort to change the task (Morin, Samozino et al. 2007) can modulate lower limb stiffness. Hobara et al (2007) used conscious effort to increase stiffness during upright hopping by asking participants to hop with reduced ground contact time. It was hypothesised (Hobara, Kanosue et al. 2007) that there would be a concurrent increase in ankle muscle co-activation with increased ankle stiffness. However, no noticeable increase in feed-forward levels of activation of the Tibialis Anterior was observed (Hobara, Kanosue et al. 2007). Thus, they concluded that individuals do not utilise a co-activation strategy to consciously control modulation of lower limb stiffness. This literature therefore establishes a hypothesis that conscious stiffness modulation via co-activation may not be the optimal control strategy for submaximal consciously controlled internal challenge.

The knowledge or expectation of pending joint loading alters motor control strategies. For example, the expectation of forces associated with landing significantly influences the muscle activity profiles compared to unexpected landings (Santello 2005). These expectations however are based on the interpretation of the external environment and the

subsequent landing perturbation (Taube, Leukel et al. 2012). In the case of consciously controlled leg stiffness modulation such as hopping with reduced contact time (Hobara, Kanosue et al. 2007) the individual is in control of the interaction with the environment. To date no studies have examined motor control strategies driving consciously (internally) controlled stiffness modulation during repeated SSC. Therefore, this paper aims to investigate kinematic and EMG signal outputs during the contact phase of two sustained submaximal hopping conditions - preferred contact time (self-selected stiffness) and shortened contact time (increased stiffness). The researchers employed double leg hopping on a custom built sleigh in order to reduce the impact of vestibular and visual input and to control for confounding factors such as fatigue and balance (Kramer, Ritzmann et al. 2010; Furlong and Harrison 2013).

The purpose of the study was firstly, to document the kinematic and muscle activity profile changes associated with conscious controlled increase of leg stiffness during repeated SSC and secondly, to investigate whether these changes were influenced by leg dominance.

Methods

Participants

This study utilised a within-subject experimental design. Following informed consent (HREC approval #PT0189) participants attended one testing session at the Institutional Motion Analysis Laboratory. Nine healthy participants (4 male, 5 female; mean (SD) age 32 (2.5), height 173.4 (9.1) cm, body mass 72.9 (14.7) kg) were tested. Inclusion criteria required that they were free of any pain or functional limitations in order to participate.

Sleigh

A custom-built sleigh apparatus with an instrumented (AMTI forceplate 1kHz sampling) landing platform was used to establish event markers during hopping e.g. contact and flight phases.

Procedure

Each trial block involved a minimum 10 continuous bilateral hops on the sleigh apparatus (median 6 used for analysis) that entailed minimising any associated knee flexion (no external fixation was used). Participants had their eyes closed and wore headphones with background music to eliminate visual and auditory feedback. Following 5 minutes of familiarisation participants performed 3 blocks of 10 hops under each of two different conditions: Preferred Contact Time (PC) using their natural ground contact time as determined during the familiarisation period and Short Contact Time (SC): with as short a contact time as possible. The order of testing was randomised for each condition and a 2 minute rest period was provided between each trial.

Fourteen Vicon (Oxford metrics, Inc.) infra-red cameras, sampling at 250Hz, captured the movement of reflective markers that were applied to the lower limbs in accordance with a cluster based model (Besier, Sturnieks et al. 2003). Reflective markers were attached to anatomical landmarks around the pelvis, thigh, leg and foot (anterior superior iliac spine, iliac crest, iliotibial band, tibial shaft, calcaneus, and 1st and 5th metatarsal heads). Leg dominance was nominated by the participant as the foot they would normally use to kick a ball (Witvrouw, Danneels et al. 2003). A validated anatomical marker set (with reconstruction errors of < 1 mm (Ehara, Fujimoto et al. 1995; Richards 1999)) and model (Besier, Sturnieks et al. 2003) was used to generate a 3D, anatomically relevant, reconstruction of the lower limb, including a foot, leg, and thigh segment. The validated mathematical model (International Society of Biomechanics, (Wu, Siegler et al. 2002) following ZXY order of rotations (Grood and Suntay 1983) generated sagittal plane ankle range of motion, position of the centre of pelvis, functional leg length and ankle joint angular displacement during hopping.

Electromyography (EMG)

An AMT-8 (Bortec Biomedical Ltd.) system was utilised to collect surface EMG signals bilaterally from the medial gastrocnemius (MG), soleus (SOL), and the tibialis anterior (TibAnt) muscles. Bipolar differential surface electrodes (Ag / AgCL) were placed on the belly of each muscle with the reference electrode on the tibial plateau. Skin impedance (< 15 kOhms) was achieved with skin preparation and signals were pre-amplified, analogue filtered (10 – 500Hz band pass) and then digitised using an 18 bit A-D card with a sampling rate of 1000Hz. All data was temporally synchronised and recorded to hard disc on the

Motion Laboratory dedicated hardware running a customised Labview program (National Instruments Inc.).

EMG signal Onsets, Normalisation and Conditioning

The EMG signal was full wave rectified and onsets detected using the integrated protocol (Allison 2003). Trial linear envelopes (LE) were created using a fourth-order, zero-lag Butterworth low-pass filter (10 Hz) and temporally synchronised to (T=0) foot contact. Ensemble average LE were determined for a 760 ms window defined as 280 ms prior to contact and 480 ms after contact. This is consistent to detect medial gastrocnemius pre-activation onsets (Jones and Watt 1971). The feed-forward threshold was defined at 33 ms after contact for both hopping conditions as this has been identified as the short latency reflex window for the soleus muscle during double leg hopping across multiple loading parameters (Voigt, Dyhre et al. 1998). EMG signals were integrated in 20 ms epochs (IEMG) for the 760 ms window. EMG data was also time normalised to the duration of the contact phase for secondary analysis. Amplitude normalisation was undertaken with the median peak of the submaximal (Allison, Marshall et al. 1993) hopping familiarisation trials (10 consecutive hops) used as 1.0 arbitrary unit.

Leg Stiffness

Leg stiffness (K) was estimated using a spring-mass model (Blickhan 1989). The following formula (Dalleau, Belli et al. 2004) was used to calculate K with the flight (Tf) and contact times (Tc) determined from force plate contact and toe off data:

$$K = \frac{(M \times \Pi(t_f + t_c))}{(t_c^2 ((t_f + t_c / \Pi) - (t_c / 4)))}$$

Unit: kN.m⁻¹

Figure 2. Formula for estimating leg stiffness

Stiffness = K; M = total body mass; t_f= flight time; t_c = ground contact time

Kinematic Variables

The stretch amplitude (in degrees) was defined as the change in the ankle joint angle from landing (contact) to the most dorsiflexed point. The stretch velocity in degrees/ second ($^{\circ}/s$) was defined as the mean dorsiflexion angular velocity, equating to the ratio of the dorsiflexion amplitude to the time interval from landing to the most dorsiflexed point.

EMG signal onset and Co-activation

Co-activation was defined as the ratio of the agonist and antagonist muscle activity. In this study, the plantar flexor muscles MG and Sol were the agonists with TibAnt as the antagonist. Co-activation ratios between the MG and TibAnt, and Sol and TibAnt were assigned the label MG/TibAnt and Sol/TibAnt. The window for co-activation ratios based on IEMG was from the onset to peak muscle activation and this was used for all muscles and conditions. For the time normalised EMG data the co-activation ratios were calculated for each data point and compared across conditions.

Statistical Analysis

All Data analysis was performed using statistical software (IBM SPSS Statistics version 2: IBM Corp©). An alpha of 0.05 was used to represent statistical significance for all comparisons.

In order to confirm that participants had achieved a stiffer performance during SC in comparison to PC, a series of paired samples t-tests were utilized to determine significant differences in mean hopping frequency, contact duration, flight duration (work done) and stiffness estimates. Paired samples t-tests were also used to compare differences in relative co-activation ratios and onset times for each muscle between the hopping conditions.

Kinematic data (stretch velocity and stretch amplitude) and peak muscle activity amplitude between the PC and SC were analysed using a 2 factor ANOVA (side x condition) and any significant interactions and main effects were reported. Thus, the null hypothesis that conscious modulation of ground contact time during hopping would not affect motor performance as determined by our derived variables was tested.

A Linear Mixed Model was utilised to identify any significant difference in onset times and time to peak EMG signal amplitude for each muscle grouped for condition and side. Further, the linear mixed model was utilised to investigate any interaction between condition, side and muscle with onset time as the dependent variable. A paired samples t-test was used to compare differences in relative co-activation ratios between hopping conditions.

Results

The participants demonstrated a statistically significant decrease in contact duration and a concurrent increase in hopping frequency and stiffness when performing SC hopping in comparison to PC hopping ($p < 0.001$) (See Table 1). These changes were observed with no systematic change in work done (flight duration, $p = 0.611$).

Table 1. Differences in the kinematic derived variables between PC and SC hopping

Variable	PC mean (SD)	SC mean (SD)	Mean Difference	95% CI	p-value
Hopping Frequency	1.36 (0.15)	1.52 (0.16)*	-.17	-.21 to -.12	<.001
Contact Duration (ms)	380 (36)	293 (30)*	87	72 to 101	<.001
Flight Duration (ms)	368 (92)	372 (98)	-4	-32 to 24	.611
Stiffness (kN.m-1)	9.20 (2.58)	14.16 (3.09)*	-4.96	-5.79 to -4.14	<.001

*Denotes a significant increase between PC and SC hopping conditions; 95% CI = 95% Confidence Interval of the difference

The mean (SD) stretch amplitude for PC was 23.6 (9.6) and 24.2 (8.3) degrees for the dominant and non-dominant legs respectively. The stretch amplitude for SC was 16.9 (4.8) and 15.9 (4.8) degrees for the dominant and non-dominant legs. There was no significant interaction between side and hopping condition ($F = 1.36$, $p = .249$). There was no main effects for side ($F = .047$, $p = .829$) but a significant difference was observed for condition ($F = 16.739$, $p < .001$). When grouped for side, the mean stretch amplitude for SC was reduced by 7.49 degrees (95% 3.82 to 11.17, $p < .001$) when compared to PC.

The mean (SD) stretch velocity analysis showed no significant interaction between side and hopping condition ($F = .174$, $p = .678$). There was no main effects for side ($F = .929$, $p = .340$) or condition ($F = .401$, $p = .529$). For the PC condition dominant and non-dominant leg was 300.5 (74.8) and 294.5 (73.4) °/sec. For the stiffer hopping strategy (SC) these values were 288.9 (52.0) and 285.9 (44.4) °/sec.

The peak EMG signal amplitude for all muscles were not significantly altered with no significant interaction between side and hopping condition ($F = .182$, $p = .671$) nor main effects for side ($F = .284$, $p = .596$) or condition ($F = .690$, $p = .409$).

The time to peak EMG for individual muscles is presented in Table 2. There were no significant interactions between side and any of the factors (muscle, condition $p = .538$), nor was there a main effect for side ($p = .812$), therefore data from both sides were pooled. Time to peak muscle activity was influenced by condition and muscle.

Table 2 shows the significant reduction in the time to peak muscle activity of the agonists muscles (Sol and MG) during the increased stiffness hopping strategy. The antagonists had a similar trend but this did not reach statistical significance. This is shown in Figure 3 with the group ensemble averages.

The mean onset time for individual muscles is presented in Table 2. Sides were pooled as there was no significant interaction between side and any of the factors (muscle or condition, $p > 0.05$), nor a main effect for dominance ($p = .127$). The agonist muscles were observed to have a shift in the onset with a significantly earlier activation in the stiffer hopping pattern.

Table 2. Agonist and antagonist muscle activity onset and time to peak (ms relative to contact) for self-paced and shortened contact sleigh hopping

		Preferred Hopping (PC)	Shortened Contact (SC)	
		Mean (SD) 95% CI	Mean (SD) 95% CI	p value
		ms	ms	
Agonists				
Soleus				
	Onset	86 (30) 58 to 114	14 (35)* -7 to 36	<.001
	Time to Peak	200 (28) 184 to 216	114 (19)* 101 to 125	<.001
Medial Gastrocnemius				
	Onset	41(18) 25 to 57	-22(14)* -35 to 9	<.001
	Time to Peak	195 (28) 179 to 211	102(20)* 91 to 114	<.001
Antagonist				
Tibialis Anterior				
	Onset	80 (80) 5 to 144	46 (71) -20 to 112	.062
	Time to Peak	218 (89) 167 to 270	145 (80) 99 to 191	.077

*Denotes a significant increase between PC and SC hopping conditions; 95% CI = 95% Confidence Interval of the mean. Negative values refer to duration prior contact.

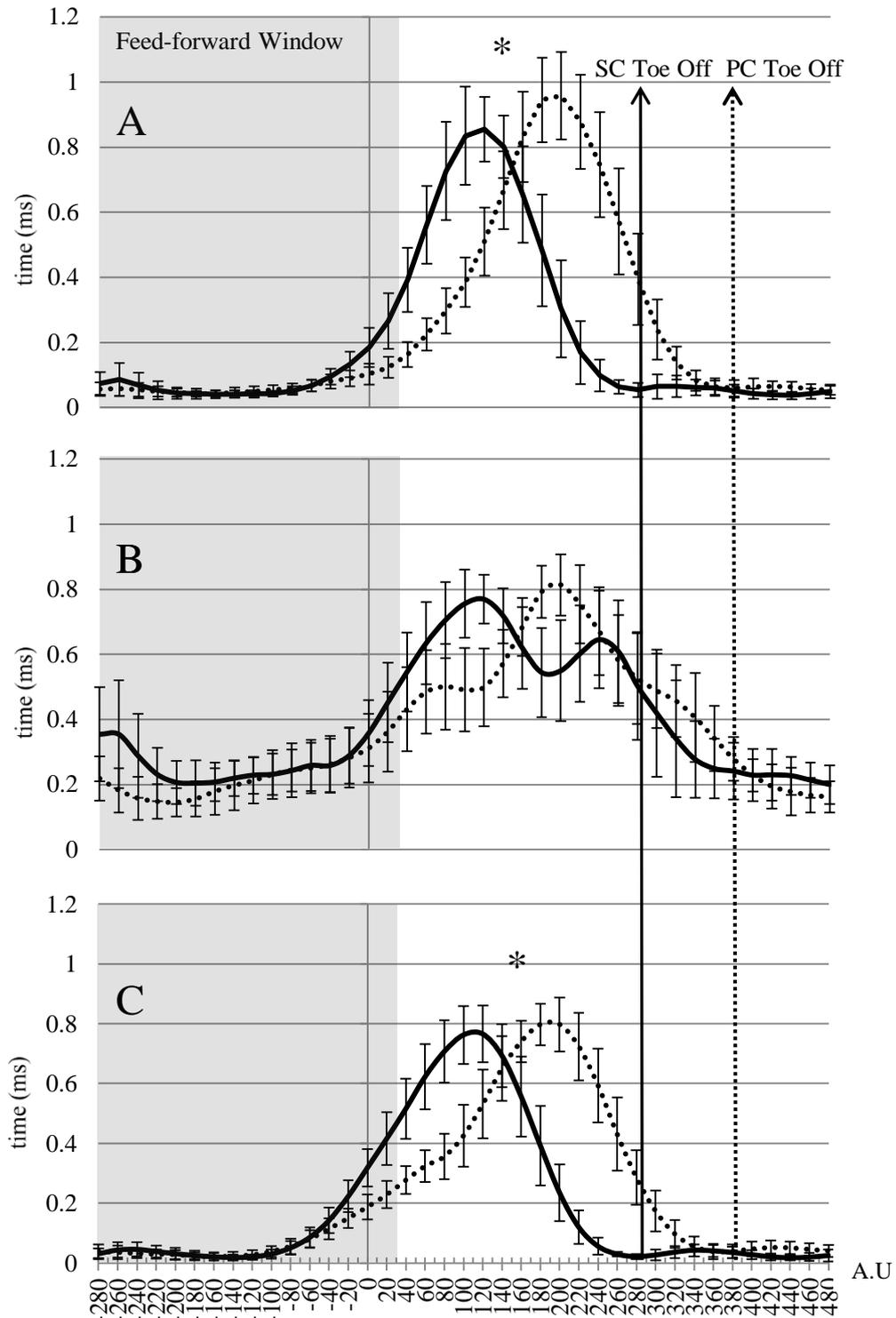


Figure 3. Mean linear envelopes for the Sol (A), TibAnt (B) and MG (C) muscles during the PC condition (dotted line) and the SC condition (solid line).

All graphs have been normalised relative to contact ($T = 0$) where minus values indicate pre-contact and the end of the contact phase has been illustrated via dotted and solid arrow for each condition respectively. *Denotes a significant difference in time to peak between PC and SC hopping conditions. A.U = Arbitrary Unit

Relative Co-activation

All comparisons revealed no significant differences in co-activation ratios for agonist and antagonists across hopping conditions ($p>0.05$). This suggests that increased leg stiffness in the SC condition was not a result of ankle co-activation. The co-activation ratios are detailed in Table 3.

Table 3. Differences in co-activation ratios between PC and SC hopping

Co-activation Ratio	Side	Condition	Mean	SD	p value
Sol / TibAnt	Dominant	PC	0.961	.167	.812
		SC	0.927	.289	
	Non-Dominant	PC	0.858	.245	.816
		SC	0.836	.241	
MG / TibAnt	Dominant	PC	0.952	.158	.483
		SC	0.871	.239	
	Non-Dominant	PC	0.675	.183	.515
		SC	0.755	.250	

Time Normalised Plots

EMG data was also time normalised to contact duration for the creation linear envelopes and co-activation profiles (Figures 4 and 5).

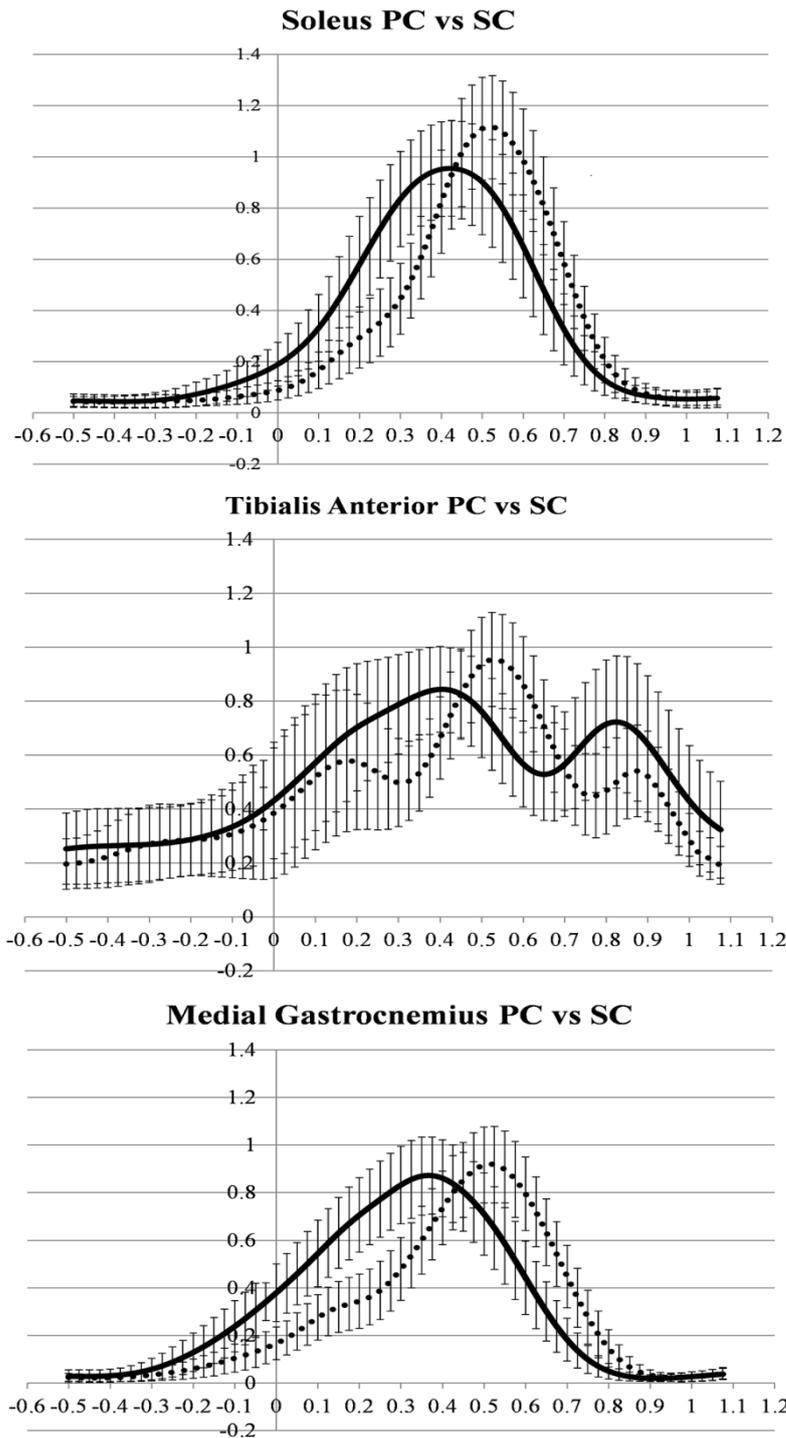


Figure 4. Time normalised mean linear envelopes for the Sol, TibAnt and MG muscles during the PC condition (dotted line) and the SC condition (solid line).

All graphs have been normalised relative to the contact phase where 0 = contact, 1 = toe-off / end of the contact phase and minus values indicate pre-contact. X-axis: time (s); Y-Axis: arbitrary unit (A.U.)

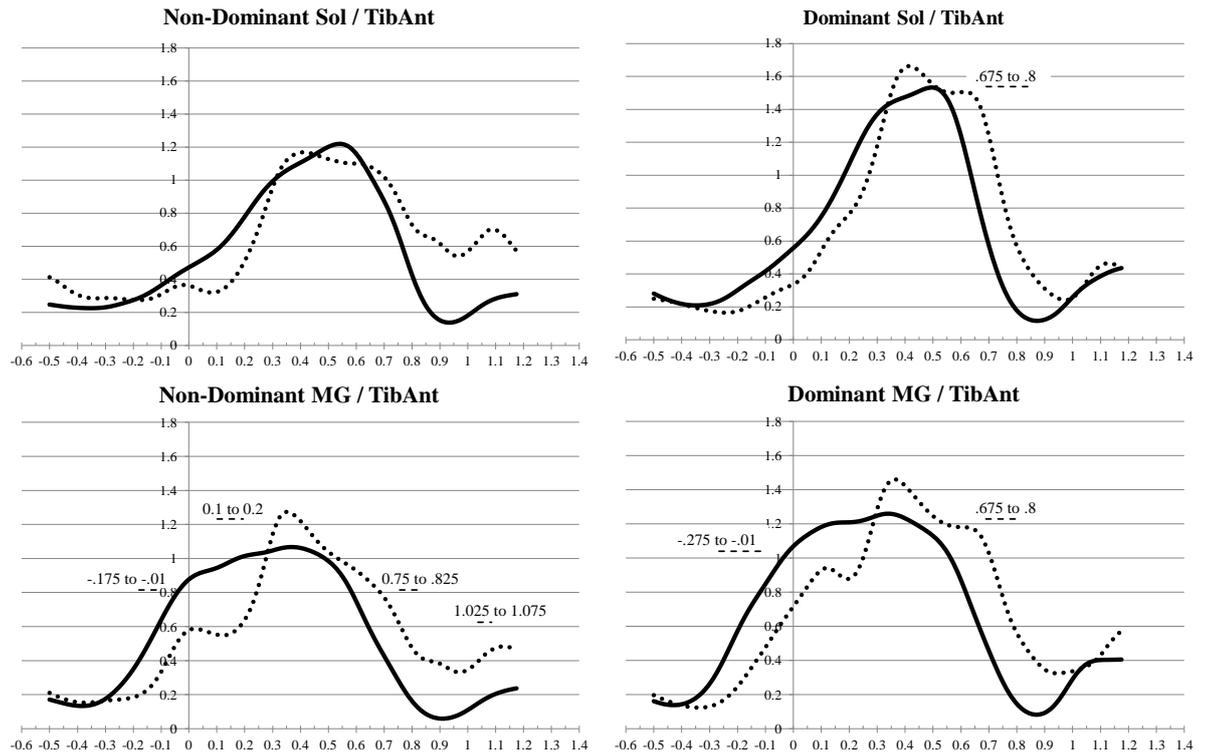


Figure 5. Time normalised co-activation profiles during the PC condition (dotted line) and the SC condition (solid line).

All graphs have been normalised to contact duration where 0 = contact, 1 = toe-off / end of the contact phase and minus values indicate pre-contact (X-Axis). ---- indicates brief periods of significant difference in co-activation ratios. Y-Axis: arbitrary unit (A.U.)

Discussion

Conscious modulation of joint stiffness has been investigated in the lower limb with the focus on changes to the external environment. This study investigated the kinematic and muscle activity profiles under two internally controlled hopping conditions that consciously altered leg stiffness in a low load environment.

When instructed to spend less time on the ground individuals responded by significantly decreasing contact time without a significant change in flight time. Since flight time on the low friction sleigh reflects the energy outputs of the system against gravity we established that the individual increased the stiffness without an increase in work. The individuals' systematic increase in the lower limb stiffness is consistent with the findings of Hobara et al (Hobara, Kanosue et al. 2007) and yet this is the first study to consider the motor control

changes with no concurrent systematic change in work done during each hop cycle between two different stiffness conditions.

The kinematic data shows that the individuals developed a strategy of reducing the range of ankle excursion during the loading phase of each hop for the SC condition. This reduction in the stretch amplitude was approximately 7.5 degrees. There was a corresponding reduction in the time to peak dorsiflexion since the rate of stretching (velocity of stretch) was not significantly altered. This clearly demonstrates that during conscious increase in joint stiffness participants relied on active elements of the system mainly dominated by muscle outputs by maintaining a mid-range position. This suggests that during consciously controlled stiffness increase participants did not rely upon the passive elements of the ankle joint which has been observed previously in response to unexpected perturbations during hopping (Moritz and Farley 2004).

Furthermore, on examining the muscle activity profiles there was no evidence that the peak EMG signal amplitudes were systematically changed between conditions. In many biomechanical models the EMG signal amplitude or drive can be used to predict increase stiffness of joints. However, this was not observed in this experiment suggesting that a 'stiffer' performance was not driven by increased muscle drive but rather an optimisation via temporal change in the muscle activation profiles, specifically a shift in timing (earlier) of the plantarflexors. This is consistent with Hof (Hof 2003) who suggested that EMG timing is more important than amplitude for modulating similar tasks. While the MG tended towards a feed-forward onset during PC hopping, both agonists (MG and Sol) had an earlier onset during SC hopping. For Sol the temporal shift in onset time represents a change from potential feedback latency to a clear feed-forward response with the onset time sitting well within the 33 (+/- 7) ms short latency reflex window identified consistently for soleus in a previous study examining double leg hopping (Voigt, Dyhre et al. 1998) (see Figure 3). We are less confident to say that the antagonist (TibAnt) had an earlier feed-forward onset ($p \approx 0.1$) since there was a wide range of onsets observed in our study population. The time to peak for the agonists was also earlier with less certain changes in the antagonist. Therefore, we observed the increased leg stiffness via a temporal shift and compression of the motor activity of the agonists during low load hopping in a controlled environment. However, it could be hypothesised that the observed temporal shift observed in Figure 3 was an anomaly created via the reduced contact time requirements of the SC condition. However, when the same EMG data was time normalised to the contact phase to control for the temporal compression of the SC condition we observed that the same temporal shift was evident (Figure 4).

We also examined the hypotheses that there would be increased co-activation ratios in the agonist-antagonist pairs in response to an increased stiffness demand. In a previous study (Moritz and Farley 2004) examining stiffness modulation during hopping, it has been observed that participants pre-activated most of their leg muscles when encountered by an expected change in stiffness of the floor surface – i.e. requiring an expected modulation of stiffness in response to a predicted external perturbation (Moritz and Farley 2004). In the same study, the stiffness of the floor surface was also changed unexpectedly and the participants did not pre-activate their leg muscles prior to landing (Moritz and Farley 2004). As previously stated, co-activation strategies represent an attempt to increase stiffness in order to reduce the risk associated with control decisions in an environment of potential/predictable reduced sensory input informing motor performance (Stroeve 1996). Clearly certainty regarding the performance outcome can influence the motor strategy selected to modulate stiffness. Our participants were in control of the increase in leg stiffness, as they were consciously driving the stiffer performance. Modulating their own contact time gave them control of their flight time and stiffness. They were always aware of their position in space and there were no other external challenges or perturbations to performance of the SC hopping condition. As per Hobara et al 2007, our data also failed to detect any increase in the co-activation ratios under different hopping conditions. The co-activation ratios were also derived using time normalised data in order to create a profile normalised to the duration of the contact phase (representing 100%) and allow comparison across conditions (Figure 5). Interestingly, there was a small window where the MG / TibAnt co-activation ratios increased pre-contact for both the dominant (27.5% to 10% pre-contact) and non-dominant legs (17.5% to 10% pre-contact). However, these brief windows did not incorporate the point of foot contact. Furthermore, examination of the linear envelopes also revealed these increases in the co-activation ratios were created by increases in MG activity and that TibAnt did not change, thus the increase in the ratios was not attributable to a true increase in co-activation. Therefore, we observed the mechanism behind increasing leg stiffness was a dynamic strategy pairing pre-activation with an increased rate of activity of the agonist muscle to develop force in time for contact with the surface. It may be hypothesised that the conscious control of the internal perturbation and the absence of an external perturbation negated the need for a co-activation strategy.

We suggest that there are at least two fundamental motor control strategies to modulate joint stiffness – co-activation (Blickhan 1989; Santello 2005; Lee, Rogers et al. 2006; Yeadon, King et al. 2010) and feed-forward response. Co-activation strategies may be most relevant to situations where there is risk associated with control decisions – i.e. perturbed

performance (Moritz and Farley 2004), reduced sensory input (Fu and Hui-Chan 2007) or expectation of pain (Moseley, Nicholas et al. 2004). In settings where there is a conscious drive to increase leg stiffness without external perturbation or risk the optimal strategy to increase stiffness may be governed by a feed-forward response. In our cohort the feed-forward response observed was a temporal shift (earlier onset) of the same motor strategy effectively matching the output of the plantarflexors to the consciously driven increased stiffness demand.

These results suggest that in the clinical setting, if the consequences of a pending perturbation are unknown but potentially significant then individuals may choose the co-activation strategy. However, in the presence of a controlled environment and self-regulation of the pending perturbation and consequences (i.e. the choice of hopping contact time on a stable sleigh) individuals may choose a strategy that alters the timing (feed-forward onset) of the strategy as observed in this study.

Relevance to other / future research

There are few hopping research studies that have considered the internal strategies for changes in lower limb stiffness (Hobara, Kanosue et al. 2007). Yet in the motor control literature for spinal instability and chronic pain it is an observation that individuals have increased co-activation and/ or a loss of feed-forward responses (van Dieën, Selen et al. 2003; Moseley, Nicholas et al. 2004; Hodges, van den Hoorn et al. 2009). It is an interesting observation that these clinical groups demonstrate movement patterns that in lower limb research suggests the individual may not be aware of the magnitude or timing of a pending perturbation - a co-activation strategy (Santello and McDonagh 1998; Moritz and Farley 2004). This appears to be a relationship worthy of further investigation.

Conclusion

This study investigated the neural control of consciously controlled increase in joint stiffness during human hopping. Leg stiffness was increased by a temporal compression of the motor pattern of the ankle plantarflexors in order to drive a stiffer hopping performance. Importantly, increased co-activation of the ankle muscles was not observed and was not a strategy for stiffness modulation during consciously controlled stiffness changes. These

findings contribute to lower limb stiffness modulation theories and are also consistent with the observed motor control strategies in some spinal pain syndromes. Therefore we conclude that there are at least two fundamental motor control strategies to modulate joint stiffness and that (as observed on spinal pain models) in the absence of co-activation strategies a preferred control mechanism is temporal modulation of the feed forward responses.

12. Supplementary Chapter 3: Do kinematic derived measures of event markers correlate with those derived using a force-plate during human hopping?

Introduction

Traditionally, force plates represent the gold standard for identifying event markers in gait analysis (Cavanagh and LaFortune 1980; Buczek, Cavanagh et al. 1991; O' Riley, Dicharry et al. 2008). As such, they are employed to establish temporal data associated with human movement e.g. stance phase or swing phase in gait. This has been transferred to analysis of human hopping where the phases of gait are paralleled by the contact and flight phases of hopping (Brughelli and Cronin 2008; Yen, Auyang et al. 2009; Bobbert and Richard 2011). The purpose of this chapter is to investigate the novel idea of defining event markers during human hopping using kinematic data. This is intended to substantiate some of the research methods of the thesis. However, there is application to other research where force plate data may be contaminated e.g. when ambulating with walking aids or where force plate data is simply unavailable. Specifically, this chapter aims to answer the following research question:

- Do Kinematic derived Measures of Event Markers correlate with those derived using a Force-plate during human hopping?

Background

The Problem

Accurate measurement of Contact and Flight Times when making changes to force plate height during an active hopping trial

Stiffness refers to the resistance of a body to a change in length (Blickhan 1989; Brughelli and Cronin 2008; Bobbert and Richard 2011). Estimating stiffness utilises a spring-mass model which represents the hopping leg as a spring which compresses and recoils under the centre of mass (Blickhan 1989; Dalleau, Belli et al. 2004). Stiffness estimates (K) are central to the current thesis line. The stiffness of the lower limb can be estimated from hopping trials using the following formula (Dalleau, Belli et al. 2004):

$$K = \frac{(M \times \Pi(t_f + t_c))}{(t_c^2 ((t_f + t_c / \Pi) - (t_c / 4)))}$$

Unit: kN.m⁻¹

Figure 1. Formula for estimating leg stiffness

Stiffness = K; M = total body mass; t_f = flight time; t_c = ground contact time

From the above formula, it is clear that accurate measures of Flight Time (t_f) and Contact Time (t_c) are critical to accurate stiffness estimate. Normally, kinetic data provides a very accurate and reliable measurement of t_c and t_f and thus it is an ideal method for defining hopping cycles. However, as outlined in the Chapter 4, the researchers constructed a Sliding Floor Mechanism (see fig 2 below) and placed it on the force plate on the inclined sleigh apparatus. The rig allowed for convenient and independent adjustment of the floor height under either foot. Although this represented an ideal method for adjusting the floor height during a hopping trial, the interface between the force plate, adjustable wooden rig and participants' foot rendered accurate measurement of hopping contact and flight phases with the force plate impossible. Since this sleigh hopping methods is used in other aspects of the research program an alternative method of assessing lower limb stiffness was required.



Figure 2. Sliding floor mechanism demonstrating a 36mm difference in floor height between left and right legs

Timing of the contact phase

The MPD protocol required participants to perform multiple trials of five hops on the inclined sleigh apparatus. During each trial of 5 hops, as the participant was mid-flight the tester made one random adjustment of the floor height under the allocated foot. However,

during piloting the movement of the boards on the force plate often created a force output in the vertical direction. Thus, the tester could unknowingly cause the force plate to register a false 'contact' or commence a legitimate contact phase unduly early. Therefore, any derived t_c and t_f could be inaccurate. Visual inspection confirmed inaccuracies in event identification. The flow on effect of such inaccuracies yielded highly inaccurate K estimates. It was therefore necessary for the researchers to find an alternative method for measuring the t_c and t_f for all studies using the Sliding Board Mechanism. The use of Kinematic data to determine the contact and flight phases in hopping represented a novel approach to this problem. For this purpose, the differences in event markers calculated with the force plate were compared with those determined from kinematic measures.

Methods

Subjects

The data from six subjects (3 female, 3 male; mean age 27; mean height 172.3cm; mean weight 73.6kg; all right foot dominant) performing one trial of 30 seconds hopping on their non-dominant foot on the inclined sleigh apparatus was utilised in this investigation. They did this on two occasions, one week apart.

The Sleigh

This study utilised a custom built sleigh apparatus as pictured in Figure 2. The sleigh was instrumented with an AMTI force plate sampling at 1000Hz as a landing platform. This force plate allowed for the establishment of event markers during hopping e.g. contact and flight phases.

3-D motion Analysis

Fourteen Vicon (Oxford metrics, Inc.) infra-red cameras, sampling at 250Hz, captured the movement of reflective markers that were applied to the lower limbs in accordance with a cluster based model (Besier, Sturnieks et al. 2003). This system has been demonstrated to be a gold standard measurement system, with reconstruction errors of < 1 mm (Ehara, Fujimoto et al. 1995; Richards 1999). A validated anatomical marker set and model (Besier, Sturnieks et al. 2003) was utilised to generate a 3D, anatomically relevant, reconstruction of the lower

limb, including a foot, leg, and thigh segment. The validated mathematical model was implemented to calculate sagittal plane ankle range of motion, position of the centre of pelvis, functional leg length and ankle joint angular displacement during hopping. This model followed the International Society of Biomechanics recommended approach to joint modelling (Wu, Siegler et al. 2002), with all angles calculated using the anatomically relevant ZXY order of rotations (Grood and Suntay 1983). Subjects had reflective markers attached to anatomical landmarks around the pelvis, thigh, leg and foot (anterior superior iliac spine, iliac crest, iliotibial band, tibial shaft, calcaneus, and 1st and 5th metatarsal heads).

The investigation of this problem will be presented in 3 stages:

1. Identifying the best kinematic output to derive t_c and t_f by comparison to F_z as the gold standard
2. Comparison of hopping phase durations between kinematic and force-plate measurement during sleigh hopping
3. Efficacy of stiffness measures determined with kinematic phases in comparison with the gold standard measure (F_z).

Stage 1: Identifying hopping event markers using kinematic and kinetic data

Take-off and landing as per the kinetic output (gold standard).

The vertical component of the ground reaction force (F_z) is the established method for identifying whether or not the foot has made contact with the force plate. Therefore, flight occurs when the F_z output reads zero. Correspondingly, the foot is in contact when the F_z output is less than zero as forces in that direction are assigned a minus value.

Kinetic Definition: Flight occurs when F_z Output = 0.

Kinetic Definition: Contact occurs when F_z Output < 0.

The kinetic and kinematic outputs were examined for all hopping trials. The researchers sought to identify the kinematic variables which consistently corresponded to the initiation of contact and flight phases as determined by the F_z output.

Determining the kinematics output that would most consistently match the Fz events

In order to utilise the kinematic data to define the events during hopping the metatarsal phalangeal marker no.1 (MTP 1) was chosen as an appropriate marker to track the movement of the forefoot. The MTP1 marker was located over the first metatarsophalangeal joint as per the description in Chapter 3.

Analysis of multiple trials utilising MTP1 position, velocity and acceleration identified the Peak Positive MTP1 acceleration as a point which corresponded to the onset of the landing phase as per the Fz output. Therefore, this was utilised to represent landing as an event marker for kinematic based outputs of t_c and t_f .

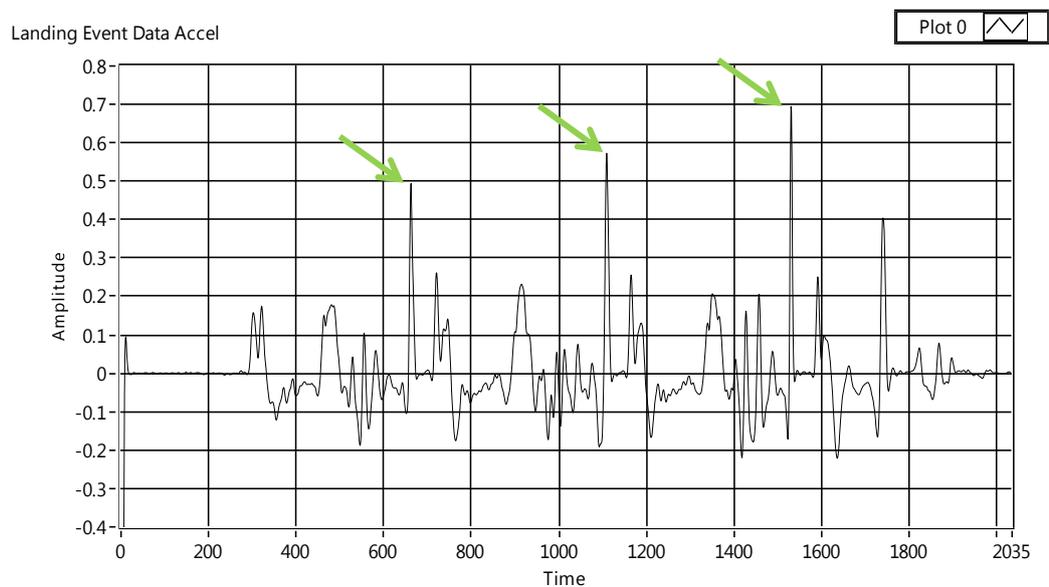


Figure 3. A graph illustrating the Peak Positive MTP1 acceleration in 3 successive hops

Take-off as defined by kinematic data?

Observational analysis of multiple trials utilising the angular velocity of the ankle joint in the X direction / perpendicular to the force plate (**AnkleX***) was identified as a variable with a specific reference point which corresponded to take-off as per the Fz output. The point at which AnkleX* crossed zero with a positive gradient after the MTP1 marker position (in mm) started increasing was utilised to represent take-off as an event marker for kinematic based outputs of flight time t_c and t_f . Put simply, the video analysis reference correlated with

toe off from the force plate data. This was represented by the change in ankle velocity from plantarflexion to dorsiflexion, after the marker at the 1st metatarso-phalangeal joint began to move away from the plate perpendicularly.

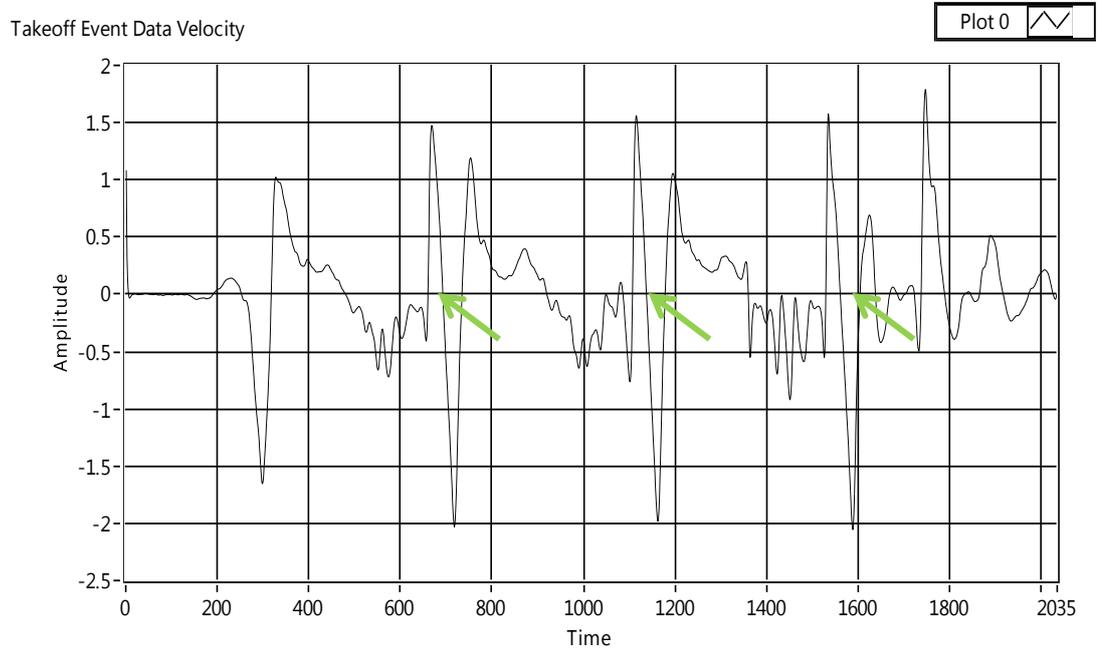


Figure 4. A graph illustrating the point at which AnkleX* crossed zero with a positive gradient after the MTP1 marker position started increasing

The following figure illustrates the temporal consistency between Fz and the kinematic variables at take-off during hopping of one participant. The broken green line signifies take off. This occurs at

1. The point at which AnkleX* (deg/s) crossed zero with a positive gradient (Upper Graph)
2. After the MTP1 marker position (mm) started increasing (Middle Graph)
3. Flight occurs when Fz Output = 0N (Lowest Graph)

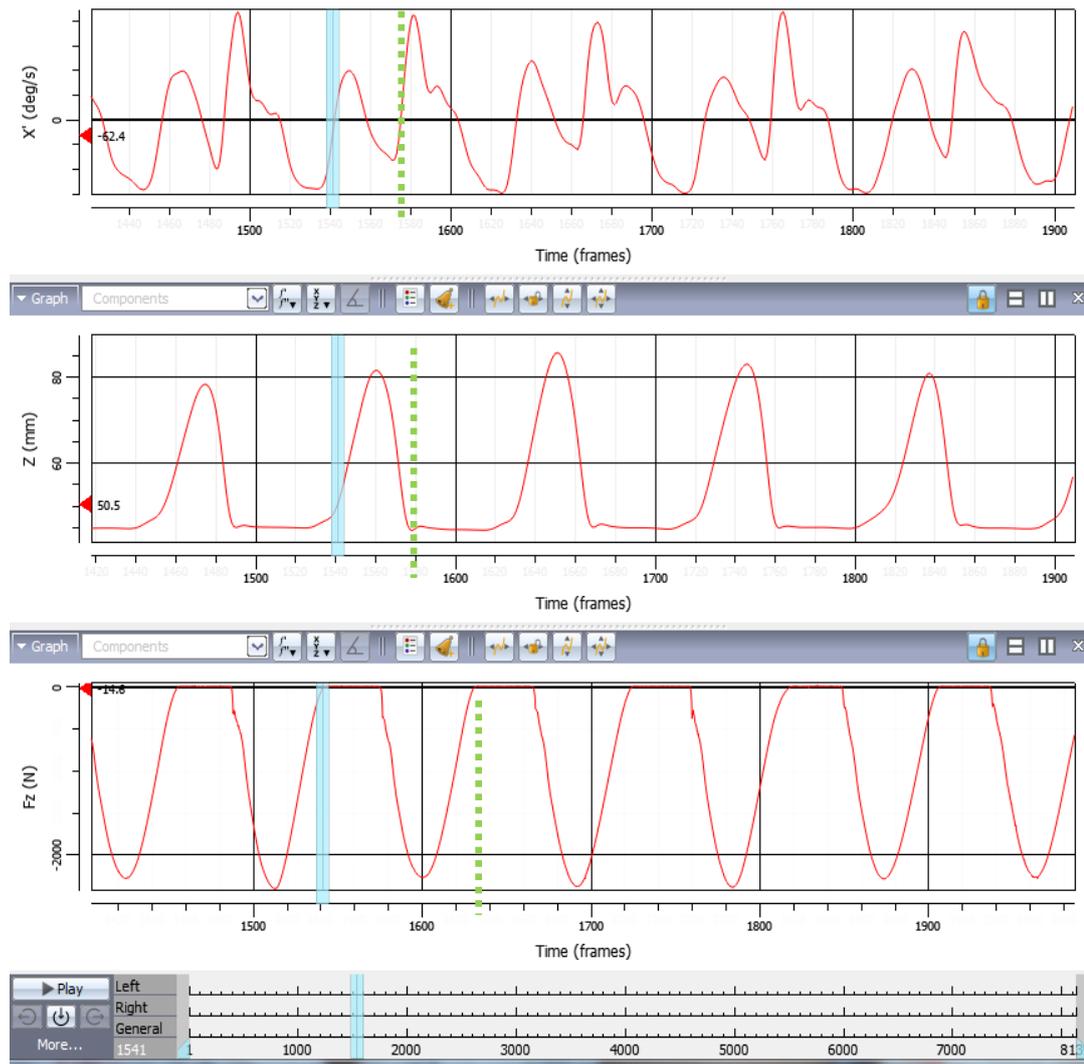


Figure 5. The relationship between the kinetic (Fz) and kinematic variables pertaining to take-off during hopping

Note: The broken green line signifies take off

Stage 2: Comparison of hopping phase duration determined by kinematics and sleigh hopping

The kinetic and kinematic measurements of t_c and t_f were compared using a dependent t-test to identify any significant differences. They were also examined for reliability using Intraclass Correlation Co-efficient (ICC) with 95% confidence intervals (CI) and Pearson r. Finally any systematic differences in t_c and t_f were investigated for systematic bias.

Results

Contact Time – Significant Difference

A significant difference was found between the force plate contact time (Ft_c) and the camera contact time (Ct_c) ($p < .0001$) (see table1). The mean difference in between Ft_c and Ct_c was found to be $-0.0246s$ (Lower 95% confidence interval limit of -0.0300 ; Upper 95% confidence interval limit of -0.0193). This was suggestive of a systematic bias in the Ct_c equating to a mean of $0.0246s$.

Table 1. Mean flight time and contact time measures from six participants performing one trial of 30 seconds hopping on their dominant foot on the inclined sleigh apparatus over two occasions

	Force Plate Data		Kinematic Data		Difference	
	Ft_c (s)	Ft_f (s)	Ct_c (s)	Ct_f (s)	t_c (s)	t_f (s)
Mean	0.4321*	0.2201†	0.4074*	0.2449†	0.0246	0.0248

*Denotes significant difference between contact times (t_c) as measured by kinetic and kinematic outputs ($p < .0001$). † Denotes significant difference between flight times (t_f) as measured by kinetic and kinematic outputs ($p < .0001$).

Contact Time – Reliability

The data was tested for reliability between kinematic and kinetic measurement of t_c and the Pearson r and Intraclass Correlation Coefficients (ICC) with upper and lower bound 95% confidence intervals (CI) were derived. A Pearson r of 0.858 and an ICC 0.814 (Lower 95% CI 0.779; Upper Bound 95% CI .844) were yielded.

An ICC of over 0.8 is considered a strong correlation and suggestive of good reliability between Ft_c and Ct_c (Chinn 1991).

Flight Time – Significant Difference

A significant difference was found between Ft_f and Ct_f ($p < .0001$) (see table1). The mean difference was found to be $-0.0248s$ (Lower 95% CI -0.0301 ; Upper 95% CI -0.0195). This was suggestive of a systematic bias in the Ct_f equating to a mean decrease in duration of $0.0246s$. Critically, this was consistent, but opposite to the systematic bias in the t_c data, see Table 2 below.

Table 2. Difference in mean t_f and mean t_c between kinematic and kinetic measurement including upper and lower bound confidence intervals

	t_c (s)	t_f (s)
Change in mean	-0.0246	0.0248
Lower 95% CI	-0.0300	0.0195
Upper 95% CI	-0.0193	0.0301

Note: CI = Confidence Interval; ICC = Intraclass Correlation Coefficient

Flight Time – Reliability

The reliability measures for the t_f data was lower than those for the t_c data, see Table 3 below. This was due to the larger relative difference in flight times due to their shorter relative durations. Nonetheless, the t_f data was deemed reliable.

Table 3. Reliability outputs for t_c and t_f Data

Reliability Measure	t_c Data	t_f Data
Pearson r	0.858	0.644
ICC	0.814	0.582
Lower 95% CI	0.779	0.514
Upper 95% CI	0.844	0.642

Note: CI = Confidence Interval; ICC = Intraclass Correlation Coefficient

The above results demonstrate that the Vicon 3-D motion analysis system reliably measured flight and contact times. Importantly, a significant difference in measures of contact time and flight time as derived by kinematic and kinetic data was observed. In particular, Table 3 illustrates a concordant yet systematic bias in Ct_c and Ct_f . The bias was almost exactly equal and opposite for Ct_c and Ct_f . This suggested that determining the hopping phase duration via kinematics resulted in a systematic overestimation of the time of contact by an average of

0.0246s and concordantly the flight time was underestimated by an average 0.0248s. The effect of this bias on Stiffness Estimates (K) will be discussed in the next section.

Stage 3: Examination of the effect of systematic bias in kinematic measurement of hopping phase duration on Stiffness Estimates

In order to outline the effect of the above described systematic bias, stiffness was estimated from both the kinematic measures and Fz. The correlation between CK and FK were examined using Pearson r and ICC with 95 % CI ranges were obtained also.

Results

Stiffness Estimates – Significant Difference

Below, Table 4 illustrates the resulting systematic bias between the kinematic derived stiffness estimate (CK) and the kinetic derived stiffness estimates (FK). The mean difference between CK and FK was -0.1947N (Lower Bound 95% CI -0.3322N; Upper Bound 95% CI -0.0573N). This equated to a 4.3% (Lower Bound 95% CI 5.5N; Upper Bound 95% CI 2.9N) overestimation of CK.

Table 4. Difference in mean kinematic derived stiffness estimate (CK) and the mean kinetic derived stiffness estimates (FK) with Standard Deviation (SD)

	CK (kN.m-1)	FK (kN.m-1)
Mean	12.859	12.664
SD	4.330	5.181

Stiffness Estimates – Correlation

The data was tested for reliability between CK and FK and the Pearson r and Intraclass Correlation Coefficients (ICC) derived. Table 5 contains reliability measurements for CK and FK.

Table 5. Reliability comparison of CK and FK

Pearson r	ICC	Lower 95% CI	Upper 95% CI
0.971	0.956	0.946	0.963

Note: CI = Confidence Interval; ICC = Intraclass Correlation Coefficient

An ICC of over 0.8 is considered a strong correlation and suggestive of good reliability between Ct_c and Ct_f (Chinn 1991). Therefore despite the bias above, CK and FL were highly correlated.

Discussion

This study identified the most appropriate kinematic measures from the foot to detect gait/hopping cycle events. The efficacy/validity of these events were compared with the known gold standard (Fz) and demonstrated to be very reliable, yet significantly different. This difference was systematic and demonstrated to have no effect on stiffness estimates. This was confirmed by both a very high reliability (ICC = 0.96) between stiffness estimated with the two different event identification methods. Further, a small systematic difference in stiffness estimates was observed, equating to a 4.3% overestimation of stiffness when using the Vicon 3-D Motion Analysis System to establish the phases of hopping.

This discrepancy was due to a 0.025 ms over-estimation of the flight time which was matched by a concordant 0.025 ms under-estimation of contact time by the Vicon 3-D Motion Analysis System when compared to the AMTI Force Plate. This may be best explained by the default 25N force-plate cut off threshold (AMTI specifications document). This is designed to control for the inevitable small fluctuations in the ambient noise which would constantly fluctuate around the zero reading of the forceplate. Without a threshold cut off these zero unavoidable crossings would be interpreted as contact phases occurring hundreds of times per second. Therefore, the true zero load is actually replaced by a 25N cut-off in order to prevent small contaminating force such as background vibrations from tainting otherwise clean kinetic data.

The diagram below (Figure 6) illustrates this force plate cut off on the measurement of the phases during hopping. This appears to be normal occurrence in kinetic measurement of hopping cycles. Thus, the 4.3% systematic difference in stiffness estimates observed above does not reflect a random error associated with using the kinematic derivation of hopping events and should be accounted for when drawing inference to existing literature. Therefore, the researchers were satisfied with the use of kinematic data to derive t_c and t_f in sleigh hopping use these outputs to estimate K as the above results demonstrate that the Vicon 3-D Motion Analysis System measured these events without the standard cut-off that occurs due to force plate threshold cut off.

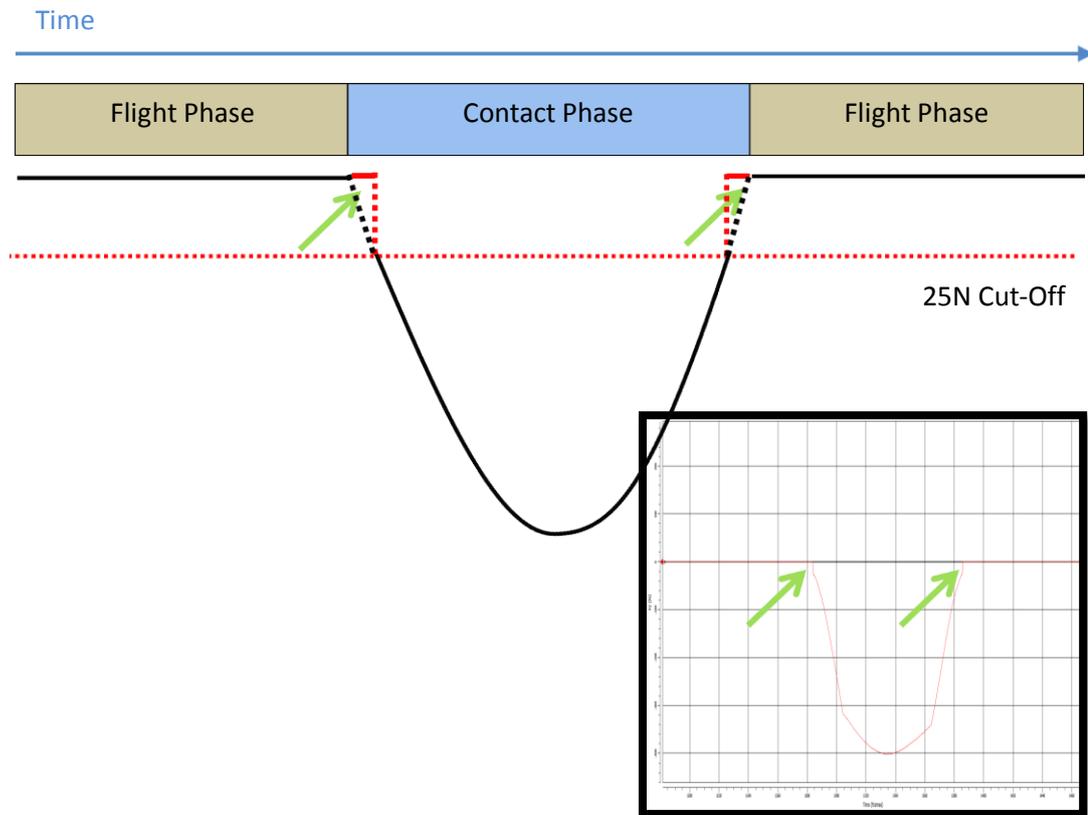


Figure 6. A schematic representation of the effect of force plate cut off on the measurement of the phases during hopping.

Note: The green arrows indicate the temporal discrepancy associated with the cut-off.

Conclusion

The use of kinematic data to measure temporal events in hopping represents a novel research method. It was trialled to overcome the issue of force plate data artefact caused by placing a sliding floor mechanism on top of the force plate as part of research protocols to test perception of changing floor high during hopping. The data collected demonstrates that the Vicon 3-D Motion Analysis System is highly accurate and reliable for measuring flight times contact times during sleigh hopping. Furthermore, a small and systematic bias in the duration of these hopping events may be accounted for by force-plate threshold cut-off.

13. References

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14. Appendices

Appendix 1	A Novel Sledge-Jump System that is Reliable for Measuring the Motor Correlates of the Stretch-Shortening Cycle (Submitted Manuscript)
Appendix 2	Participant information and consent form 1
Appendix 3	Participant information and consent form 2
Appendix 4	Participant information and consent form 3
Appendix 5	Travers M, Debenham J, Gibson W, Campbell A, & Allison G (2013) If proprioception is relevant to locomotion, then why test it standing still? 1 st International Congress on Sports Sciences Research and Technology Support (ICSPORTS), Villamoura, Portugal, September.
Appendix 6	Travers M, Debenham J, Gibson W, Campbell A, & Allison G (2013) Concious drive to stiffen the leg spring – Concious drive for an internal challenge. 1 st International Congress on Sports Sciences Research and Technology Support (ICSPORTS), Villamoura, Portugal, September.

Appendix 1

Authors

James Debenham ^{#1, 2}

Mervyn Travers ¹

Dr Amity Campbell ¹

Dr William Gibson ^{1,2}

Professor Max Bulsara ²

Professor Garry T Allison ¹

Title

A Novel Sledge-Jump System that is Reliable for Measuring the Motor Correlates of the Stretch-Shortening Cycle

Affiliations

¹ School of Physiotherapy, Curtin University, Kent Street, Bentley, Western Australia, 6102

² School of Physiotherapy, University of Notre Dame Australia, 19 Mouat Street, Fremantle, Western Australia, 6959

Corresponding Author

James R Debenham- james.debenham@nd.edu.au; Tel +61(8)9433 0996; Fax +61(8)9433 0210

Abstract

Purpose: Understanding the stretch-shortening cycle (SSC) may assist clinicians identify factors that influence lower limb injury. This study quantified the reliability of motor correlates of the SSC using a novel sledge-jump system (SJS).

Methods: A 3D motion analysis system recorded lower limb kinematics on fifteen healthy participants over two separate occasions performing 10 x 30 second trials of single-leg sub-maximal hops on a SJS. Lower limb stiffness, ankle angle 80 ms prior to contact, ankle angle at contact, ankle angle at take-off, and stretch amplitudes at the ankle, knee and hip were analysed for temporal reliability. Comparisons using a three-level intraclass correlation coefficient model were made between 3 hopping periods of 10 hops, between multiple trials and between 2 hopping occasions.

Results: With the exception of knee stretch amplitude, no significant differences were present between hopping periods ($p < 0.05$). Across trials, lower limb stiffness, ankle angle 80 ms pre-contact, ankle angle at contact and ankle stretch amplitude all demonstrated strong reliability (ICC's = 0.77, 0.86 and 0.71 respectively); ankle angle at take-off demonstrated moderate reliability (ICC = 0.68); knee stretch amplitude and hip stretch amplitude demonstrated poor reliability (ICC = 0.11 and 0.28 respectively). Testing occasion had minimal influence on measures that had moderate to strong inter-trial reliability and a moderate influence on measures that had poor inter-trial reliability.

Conclusion: A controlled assessment protocol produces reliable results in healthy adults. As such, the use of the described SJS is a reliable method of investigating biomechanical measures of the SSC.

Key Words

Spring mass model

Stretch-shortening cycle

Hopping

Lower limb stiffness

Musculotendinous mechanics

Kinematics

Introduction

Understanding the mechanical behaviour of the lower limb during functional tasks is important to optimise athletic performance, minimise injury risk and improve rehabilitation strategies (Hobara et al. 2012c). A key feature of lower limb function is the utilisation of the stretch-shortening cycle (SSC) observed during musculotendinous activation, stretch and immediate shortening. This mechanism optimises efficiency during locomotion (Walshe et al. 1998) and is a key feature of the spring mass model that describes human locomotion based on a point mass supported by a single, 'massless' linear 'leg spring' (Le Meur et al. 2012). In this model, lower limb stiffness refers to the ratio between force and linear displacement of the lower limb and is highly representative of the mechanics of running (Blickhan 1989). With consideration of the SSC in mind, researchers have moved away from evaluating motor behaviour using isometric, concentric and eccentric contractions in isolation, in favour of evaluating the SSC due to this functional utility (Komi 2000; Cormie et al. 2010). Measurements of SSC performance and its modulation may provide insight into the biomechanical aetiology of lower limb musculoskeletal pathologies and the effect of exercise based intervention strategies (Harrison et al. 2004; Hobara et al. 2012a; Maquirriain 2012).

To date, the most common SSC tasks investigated have been walking, running and hopping (Brughelli and Cronin 2008). Walking and running inherently have significant variability in performance secondary to factors such as environment, surface, balance and fatigue (Furlong and Harrison 2013). In addition, it is difficult to isolate single joints during walking/running; due to this, hopping is often employed as a surrogate model for repetitive SSC tasks with minimal extraneous factors to confound results (Farley and Morgenroth 1999; Funase

et al. 2001; Günther and Blickhan 2002; Chelly and Denis 2002; Moritz and Farley 2004; McLachlan et al. 2006; Hobara et al. 2009; Oliver and Smith 2010; Bobbert and Casius 2011; Hobara et al. 2012b; Joseph et al. 2012; Haeufle et al. 2012). Hopping is thought to have improved reliability compared to running and walking because of the restricted movement of the centre of mass and reduced degrees of freedom in the lower limb joints (McLachlan et al. 2006). However, employing an upright hopping model may still demonstrate variability in performance due to difficulty in completely controlling centre of mass trajectory, and also the rapid onset of fatigue (Horita et al. 1996). A sledge-jump system (SJS) offers an alternative model whereby individuals can hop, while the body mass is partially supported in order to restrict the degrees of freedom of the task and eliminate fatigue. This system therefore potentially affords optimal reliability within this dynamic environment.

A variety of different SJS's have been employed in the past (Aura and Komi 1986; Bubeck and Gollhofer 2001; Ishikawa and Komi 2004; Ertelt and Blickhan 2009; Kramer et al. 2010; Furlong and Harrison 2013) and several different technical approaches have been employed in developing SJS's. Despite hopping on a SJS being a promising methodology for investigating tasks with the SSC as an integral component, there is limited information on the reliability of the derived measurements (Harrison et al. 2004; Flanagan and Harrison 2007; Furlong and Harrison 2013). Likewise, all methodologies and measurement tools yield an outcome which is indicative of the 'true' result and that attributed to systematic natural random sources of variation in performance such as motivation, expertise and attention (Portney and Watkins 2008). It is important to quantify the reliability of the SJS system and its associated standard error of measurement (SEM) in order to be able to confidently identify changes in performance of the task secondary to experimental manipulation or when investigating the effect of pathology or exercise-based rehabilitation interventions (Joseph et al. 2012).

This study therefore aimed to detail the temporal reliability and SEM of lower limb kinematics and biomechanical measures of the SSC during sub-maximal single-limb hopping in healthy volunteers. It was hypothesised that after a period of familiarisation, lower limb stiffness and ankle kinematic measures recorded on this device would be reliable within a trial of 30 hops, across 10 hopping trials and between two identical hopping occasions, 1 week apart.

Methods

Participants

Fifteen healthy participants (8 females and 7 males; mean age 27.4 ± 5.6 years, height 171.2 ± 10.7 cm, mass 71.1 ± 15.9 kg) took part in this study. Participants were excluded if they were pregnant, had significant medical or psychological illness, significant visual impairment, or if medications that may affect motor performance were being taken. Likewise, participants were excluded if they had had significant lower quadrant neuromusculoskeletal pathology in the preceding 12 months, or if they had ever undergone lower limb surgery. The study was approved by the relevant institution human ethics committee (#PT0151) and all participants provided written informed consent prior to participation in this research. No participants were excluded or withdrew from this study.

Procedures

Participants attended two separate data collections, one week apart. They were instructed to continue with their normal everyday activities, but to refrain from undertaking any unfamiliar physical activity in the week prior to, and between testing occasions. In addition, they were not to undertake vigorous physical activity in the 24 hours prior to testing. Data collection was performed in the motion analysis laboratory at the School of Physiotherapy, Curtin University, Western Australia.

Initially, relevant demographic and anthropometric data was recorded (age, height, body mass, lower limb dominance). Retro-reflective markers were then affixed to the skin of the participants according to a customised marker set and model for the lower limbs and pelvis (see Figure 1), set according to an established cluster based method (Besier et al. 2003; Wu et al. 2002). Single markers were placed on the head of the first and fifth metatarsals, calcaneus, anterior superior iliac spines (ASIS) and posterior superior iliac spines (PSIS). Marker

clusters (three markers attached to a semi-rigid plastic base plate) were attached to the lateral aspects of both thighs and legs. The three-dimensional position of these markers was tracked using a 14-camera Vicon MX motion analysis system (Vicon, Oxford Metrics, Oxford, UK) operating at 250 Hz. Prior to dynamic trial collection, a static calibration trial was performed using a foot-calibration rig to measure foot abduction/adduction and inversion/eversion angles. The additional use of medial and lateral malleoli and medial and lateral femoral condyle marker locations determined anatomically relevant ankle, knee and hip joint axes of rotation and joint centres (Besier et al. 2003).

Ten 30 s dynamic hopping trials were then completed with a 90 s rest period between trials. The task involved continuous sub-maximal single-limb hopping on a custom-built SJS that included a low-friction sled, reclined to 20° relative to the base which was composed of a 50.2 cm x 50.2 cm force plate (AccuGait; AMTI, Watertown, MA) operated at 1000 Hz (Figure 2). The apparatus has been developed locally to quantify the mechanical properties of the lower limb musculoskeletal structures in vivo during sub-maximal dynamic exercise.

Participants were instructed to keep their non-hopping limb in a flexed position. Their foot rested on the SJS and they held onto the sliding component of the SJS in order to stabilise the thorax and upper limbs. Participants hopped on their non-dominant leg; this was defined as the side opposite to the participants preferred jumping leg as recommended by Flanagan and Harrison (2007). Whilst hopping, participants were instructed to keep the hip and knee as straight as possible, effectively isolating the task to the ankle. They were instructed to hop at a sub-maximal level, described as an effort they could maintain 'indefinitely'. Given the uniqueness of the task, participants were provided with a demonstration and familiarisation period until they could hop as instructed comfortably; this process typically took 10 minutes. All participant instructions/feedback were consistent, as different verbal cues can modulate lower limb stiffness in similar tasks (Arampatzis et al. 2001). This constituted one testing occasion, which was repeated exactly 7 days later.

Data Processing

The Vicon data was processed using Vicon Nexus motion analysis software (Vicon, Oxford Metrics, Oxford, UK). Data were filtered using a fourth-order low-pass Butterworth filter operating at a cut-off frequency of 20 Hz for the marker trajectories and 50 Hz for the ground-contact data as determined by residual analysis (Winter 1990). All lower limb anatomical and joint coordinate systems were calculated in accordance with the standards outlined by the International Standards of Biomechanics (Wu et al. 2002) and have been previously described (Besier et al. 2003).

Data was exported from Nexus for further analysis using a customised LabVIEW program (National Instruments, Austin, TX, 2011). For each hopping trial, this program calculated ankle joint stiffness, ankle dorsiflexion angle at 3 time points determined using force plate data (80ms prior to foot strike, at foot strike, and at take-off), as well as ankle, knee and hip stretch amplitude (range of excursion between contact and maximum dorsiflexion). Lower limb stiffness was calculated using the method described by Dalleau et al. (2004) (Figure 3).

A 30 s epoch was chosen for each block of hopping as pilot testing established that this would ensure at least 30 consecutive hops for all participants. Therefore, individual trials were sub-divided into 3 periods of 10 hops (1-10, 11-20 and 21-30). Individual period values were the lowest level of analysis and a median value of each period was selected for comparison.

Statistical Analysis

Reliability data analyses were performed using Stata (StataCorp. 2011. Stata Statistical Software: Release 12. College Station, TX: StataCorp LP). All other statistical analyses were conducted using SPSS 19 (SPSS, Chicago, IL, USA). Data was checked for normality and values outside of two standard deviations from the mean were considered outliers and were removed for the purposes of statistical analysis as described in previous literature employing SJS's (Ertelt and Blickhan 2009). On average, 4% of data was removed from each trial, consistent with typical data processing errors experienced using this model (Ertelt and Blickhan 2009).

Data were compared between the 3 periods, averaged across the 10 trials and 2 testing occasions, using a generalised linear mixed model. An alpha of 0.05 was utilised to represent statistical significance. Within-trial and within-week reliability was assessed by generating ICC values for each trial. Week was taken into consideration using the variance of intraclass correlation coefficients in three-level model as described by Hedges et al. (2012). In this model, reliability values are attributed to contributing sources (subject and week) for each trial. Previous research (McLachlan et al. 2006; Joseph et al. 2012), supported by our own pilot testing indicated trial 1 may demonstrate poorer stability than trials 2-10; therefore an additional pooled analysis was conducted for trials 1-10 and trials 2-10. ICC values above 0.70 were considered to represent strong reliability; values between 0.50-0.70 were considered to represent moderate reliability and values below 0.50 were considered to represent poor reliability (Portney and Watkins 2008). Finally a comparison by week was undertaken. According to the model by Hedges et al. (2012), low ICC values represent minimal influence on reliability between weeks and indicate higher reliability. Between-week consistency was also assessed by calculating the standard error of measurement (SEM) and minimal detectable difference (MDD) between weeks.

Results

Reliability across Individual Trials

The results from the within trials analysis revealed that there was no significant difference in lower limb stiffness, ankle and hip kinematics between periods (Table 1). A main effect was detected between periods for knee stretch amplitude ($p < 0.05$), with further analysis revealing periods 1, 2 and 3 were all significantly different.

Reliability across Multiple Trials

ICC values for all 7 measures are represented in Table 2 and Figures 4-6. **Lower limb stiffness** demonstrated strong reliability; lowest ICC value was for trial 1 (0.69), besides which individual trials ranged from 0.74-0.84, with pooled trials 2-10 ICC of 0.77. **Ankle angle 80 ms pre-contact** demonstrated strong reliability; lowest ICC value was trial 8 (0.81), besides which individual trials ranged from 0.83-0.87, with pooled trials 2-10 of 0.86. **Ankle angle at contact** demonstrated strong reliability; lowest ICC value was for trial 4 (0.77), besides which values for individual trials ranged from 0.78-0.88 with pooled trials 2-10 ICC of 0.83. **Ankle angle at take-off** demonstrated moderate reliability; lowest ICC value was for trial 2 (0.56), besides which ICC values for individual trials ranged from 0.64-0.79, with trials 2-10 ICC of 0.68. **Ankle stretch amplitude** demonstrated strong reliability; lowest ICC value was for trial 1 (0.38), besides which ICC values for individual trials ranged from 0.56-0.78, with trials 2-10 ICC of 0.71. **Knee stretch amplitude** demonstrated poor reliability; lowest ICC value was for trials 1-3 (0.00), besides which, ICC values for individual trials ranged from 0.05-0.30, with pooled trials 2-10 ICC of 0.11. **Hip stretch amplitude** also demonstrated poor reliability; lowest ICC value was for trial 2 (0.06), besides which individual trials ranged from 0.18-0.51, with pooled trials 2-10 ICC of 0.28.

Reliability between Weeks

The results regarding the influence of week demonstrated a weak effect on the variation within the results for lower limb stiffness (ICC trials 2-10 pooled = 0.19), ankle angle 80 ms pre-contact (ICC trials 2-10 pooled = 0.11), ankle angle at contact (ICC trials 2-10 pooled = 0.12), ankle angle at take-off (ICC trials 2-10 pooled = 0.27) and ankle stretch amplitude (ICC trials 2-10 pooled = 0.24). Week demonstrated a moderate influence on variation within the results for knee stretch amplitude (ICC trials 2-10 pooled = 0.50) and hip stretch amplitude (ICC trials 2-10 pooled = 0.52). Table 3 shows the SEM and MDD values for variables between weeks 1 and 2; these indicate that with the exception of knee stretch amplitude (SEM 5.39° and SDD 14.94°), the values were indicative of good consistency (lower limb stiffness SEM 0.42 kNm⁻¹, MDD 1.16 kNm⁻¹; ankle and hip angles SEM 0.97-1.45°, MDD 2.69-4.02°).

Discussion

This study investigated the reliability of lower limb kinematics and biomechanical measures during sub-maximal single-limb hopping on a novel SJS. To our knowledge, this is the first study of this nature and only

the second study exploring reliability of lower limb motor performance during the SSC after that of McLachlan et al. (2006). We employed a single-group repeated-measures study design in which participants hopped continuously at a sub-maximal level on a purpose-built SJS. The results of this study largely support our hypothesis that the SJS is reliable both within and between days. This is only the second study after Furlong and Harrison (2013) to formally report the temporal reliability of a SJS, and to our knowledge, the first to report reliability between trials and between testing occasions.

Our results indicate that within any given hopping trial, with the exception of knee stretch amplitude, there was no systematic variation in performance between the 3 hopping periods. From this we can conclude that high performance stability exists within each trial. Between trials and weeks, our results demonstrate strong reliability across a range of biomechanical measures during a SSC task, including lower limb stiffness, ankle angle 80 ms pre-contact, ankle angle at contact and ankle stretch amplitude. Between trials, ICC values for these parameters ranged from 0.68-0.86. Ankle angle at take-off demonstrated only moderate reliability; this may be expected as this measure is taken at the final stage of the task, where variation could be expected to be greatest as the foot leaves the ground and the task is completed (Austin et al. 2002). In further support of the study hypothesis, poor reliability for knee and hip kinematics were observed (ICC values of 0.11 and 0.28 respectively). Whilst ICC's reflect the variance across trials relative to the sample population, the absolute changes in knee stretch amplitude was actually relatively small (less than 1° for every 10 hops) and this is reflected in our SEM measurement of 5.39° between weeks.

Given the nature of the task, poor reliability of the knee and hip measures is not surprising; whilst the hip and knee were not externally constrained, participants were instructed to keep the hip and knee in as static a position as possible. The intention of this instruction was to isolate the task as much as possible to the ankle, given that lower limb stiffness is predominantly modulated by ankle stiffness (Farley and Morgenroth 1999), particularly during faster sub-maximal lower limb (e.g. running). Likewise, lower limb stiffness during such tasks is contributed to minimally by the hip and knee (Farley et al. 1998; McLachlan et al. 2006). This method is consistent with that of Joseph et al. (2012) who demonstrated good reliability of lower limb stiffness during sub-maximal upright hopping whilst constraining hip and knee motion via verbal instruction. Pilot testing demonstrated that to externally constrain the knee to a greater extent than could be done under volitional control was unfeasible and potentially unsafe. As such, a small degree of motion was expected and this was observed (mean knee stretch amplitude = 9°; mean hip stretch amplitude = 6°). Given the small ranges involved any variations between individuals and between trials would magnify poor ICC reliability. As complete isolation of the ankle was not achieved it is possible that the changes in hip and knee position may have had an influence on measures of lower limb stiffness and ankle kinematic measures (Kamiya et al. 2006; Kawashima et al. 2006; Knikou and Rymer 2002); however, given the relatively small changes in range of motion and our strong reliability measures at the ankle, it is expected that this effect would have been negligible. Finally, our results demonstrate that for our main parameters of interest, testing occasion had a minimal influence on reliability, providing confidence in the use of this methodology across repeated measures and over weekly intervals. To this extent the use of a three-level intraclass correlation coefficient model presents a novel interpretation of our data, which is in contrast to conventional interpretations of ICC data. It is important to recognise that using this model the low ICC values observed between weeks represent high levels of stability when comparing measures between weeks. This is because the value reflects the influence that week has on the stability, which our data indicate is small.

Of interest, our results demonstrate that for lower limb stiffness, ankle stretch amplitude, and knee stretch amplitude, exclusion of the first trial improves the reliability of the measure. However, for ankle angle 80 ms pre-contact, ankle angle at contact, ankle angle at take-off and hip stretch amplitude, exclusion of the first trial does not improve reliability. The exclusion of the first trial improved reliability presumably by mitigating familiarisation/learning that may have occurred with this task. This study provides the first objective evidence supporting the apparent empirically-derived practice of excluding first and final hops from hopping trials using this methodology (Joseph et al. 2012; Furlong and Harrison 2013). Given the improvements observed with this strategy, we recommend such exclusion in future studies utilising a SJS. In addition, a suggested benefit of using the SJS over upright hopping is the avoidance of fatigue that may directly influence results. The use of an

upright hopping model could conceivably result in fatigue towards the end of any given 30-hop trial, as indicated by an increase in performance variability. In using the SJS, our data (excluding knee stretch amplitude) remains stable in the final period of hopping. In the presence of discernible fatigue, one might expect to see a change in data, such as an increase in variability or a drift in performance. The absence of such patterns indicates that participants performing this task do not experience fatigue sufficient to contaminate performance.

The current findings of low SEM (relative to the overall total possible range of kinematic scores) coupled with the high reliability findings provide a stable platform upon which to base judgement about true change in outcome measures across time. This is important given that studies exploring the temporal behaviour of ankle performance are reliant on stable measures. Additionally, this current study provides the basis for using this methodology to predict true change in performance across time/repeated trials.

Whilst the use of SJS's is not new (Aura and Komi 1986; Kyröläinen and Komi 1995; Bubeck and Gollhofer 2001; Ertelt and Blickhan 2009; Kramer et al. 2010), ours is the first to report measures of temporal performance reliability. Harrison et al. (2004) investigated differences in SSC-function (including lower limb stiffness) between power and endurance athletes using a SJS. These authors reported reliability data from pilot testing 4 drop jumps on their SJS with ICC values of 0.996 for impact velocity. Likewise Flanagan and Harrison (2007) investigated SSC performance utilising a SJS; they performed a reliability analysis demonstrating strong reliability of vertical stiffness, flight time and reactive strength index ($ICC > 0.85$). Furlong and Harrison (2013) explored the within-trial reliability of an inverted SJS, finding good reliability of kinematic measures of ankle function ($ICC > 0.95$). These data are similar or stronger than our own and our findings add to the body of literature indicating SJS have good temporal reliability. The findings from our study can also be added to those of reliability studies conducted for upright hopping. McLachlan et al. (2006) compared ankle and lower limb stiffness during upright hopping over three separate occasions, using a comparable methodology to our own. The findings of our study reflect theirs; they observed good reliability of lower limb stiffness over three hopping occasions, whilst measures of ankle performance (ankle stiffness) also demonstrated good reliability if the first 'familiarisation' trial was discarded. Lloyd et al. (2009) investigated the temporal reliability of sub-maximal two-legged upright hopping, demonstrating 'acceptable' reliability of lower limb stiffness. Interestingly, their analysis included exclusion of the first trial, and they observed no noticeable improvement in reliability. This is in contrast to our own results and may reflect differences in methodology. Joseph et al. (2012) investigated the temporal reliability of lower limb stiffness, as well as hip, knee and ankle stiffness during submaximal upright hopping, observing good lower limb stiffness measures, but poor to moderate joint stiffness measures. They also observed superior reliability when the hopping frequency was forced rather than self-selected. Besides ankle stretch amplitude, our comparable results of ankle kinematics demonstrated superior reliability, indicating that notwithstanding the differences in the nature of the task (upright vs. sleigh), our method of exploring the SSC demonstrates value-add in terms of reliability.

Limitations to our study must be mentioned. Learning processes involving cerebellar and cerebrocerebellar communication loops, as well as mechanical issues such as tendon hysteresis may naturally influence performance of the SSC-task. Historically SJS's have been criticised for task unfamiliarity contributing to lack of ecological validity; we attempted to account for this by sampling a large number of hops within trials and a large number of trials (10 trials of 30 hops). This was performed in an attempt to produce a normally distributed sample and elucidate statistically significant results. This contrasts previous similar studies where typical sampling values vary from 3 trials of 20 hops (Lloyd et al. 2009) to 10 trials of 10 hops (Joseph et al. 2012). In addition, our instructions to participants indicated they should hop at a sub-maximal level, described as an effort they could maintain 'indefinitely'. Removing the subjectivity of this instruction could potentially further improve the reliability of the measure. Furlong and Harrison (2013) have identified a method of semi-quantification, where optimal reliability was achieved based on the instructions to hop at a level of 2 out of 5 on a numerical rating scale, where 5 represents the maximal force they could apply whilst still successfully completing the task. Horita et al. (1996) have attempted to quantify/define sub-maximal hopping. Using the SJS described by Aura and Komi (1986), they defined sub-maximal hopping as hopping at 70% of the maximal countermovement jump. Using this method, participants fatigued at 3 minutes of hopping on average. Given these results we can be confident that our protocol would not have resulted in significant levels of fatigue, and

this is reflected in the reliability of our results. Finally, our study required participants to hop at a self-selected pace. Whilst it has been demonstrated that hopping at a self-selected frequency results in poorer reliability than a forced frequency (Joseph et al. 2012), Dalleau et al. (2004) suggests a difference of 0.6 Hz in hopping frequency is required to significantly alter stiffness values. Participants in our study did not vary to this extent, and as such these results imply that the use of a SJS has the advantage of yielding reliable measures, whilst allowing participants to hop at a 'natural/preferred' frequency.

Conclusions

Based on the results of the current study, it is proposed that a strictly controlled assessment protocol yields outcome measures of SSC measures which demonstrate good reliability. Secondly, our novel description of SEM around motor correlates of this task within this methodology provides a firm basis upon which future methodologies utilising this protocol may (with the appropriate degree of confidence interval) infer true change across experimental conditions. As such, the use of the described SJS is an ecologically valid method of investigating the motor correlates of the SSC. It is recommended that future studies utilising a SJS employ similar methods of measurement as described here; notably, a familiarisation period is recommended where data from the initial trial is not analysed. Finally, multiple-occasion testing is an appropriate strategy and we present documented measures that reflect natural variation in the measure.

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Conflicts of Interest

None

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Tables

Table 1: Means (and standard deviations) for different periods during sleigh hopping (weeks 1&2 pooled)

Variable	Period 1	Period 2	Period 3
Lower limb stiffness (kNm ⁻¹)	9.43 (±3.86)	9.41 (±3.89)	9.44 (±3.91)
Ankle angle 80 ms pre-contact (dorsiflexion)	-19.60 (±10.48)	-19.74 (±10.41)	-20.30 (±10.65)
Ankle angle @ contact (dorsiflexion)	-21.19 (±10.47)	-21.15 (±10.24)	-20.69 (±10.20)
Ankle angle @ take-off (dorsiflexion)	-30.44 (±8.34)	-30.47 (±8.46)	-30.19 (±8.52)
Ankle stretch amplitude (degrees)	28.53 (±9.63)	28.63 (±10.24)	29.81 (±9.53)
Knee stretch amplitude (degrees)	8.99 (±4.37)*	9.78 (±4.81)*	10.59 (±5.02)*
Hip stretch amplitude (degrees)	6.30 (±2.31)	6.37 (±2.28)	6.32 (±2.28)

Table 1- Means (and standard deviations) for different periods during sleigh hopping (weeks 1&2 pooled)

* Significant difference between periods 1-3 (p < 0.05)

Table 2: Reliability of Stiffness Measures (ICC) for Trial, Session and Week

Trial	Stiffness	Ankle Angle 80 ms Pre-Contact	Ankle Angle at Contact	Ankle Angle at Take-Off	Ankle Stretch Amplitude	Knee Stretch Amplitude	Hip Stretch Amplitude
1	0.69 [0.41-0.96]	0.84 [0.70-0.99]	0.83 [0.68-0.99]	0.69 [0.42-0.95]	0.38 [0.00-0.82]	0.00 [0.00-0.00]	0.27 [0.00-0.72]
2	0.75 [0.53-0.98]	0.86 [0.73-0.99]	0.83 [0.67-0.98]	0.56 [0.20-0.91]	0.61 [0.29-0.93]	0.00 [0.00-0.00]	0.06 [0.00-0.54]
3	0.80 [0.61-0.99]	0.87 [0.74-0.99]	0.85 [0.71-0.99]	0.65 [0.36-0.95]	0.68 [0.40-0.96]	0.00 [0.00-0.00]	0.18 [0.00-0.65]
4	0.76 [0.54-0.98]	0.84 [0.68-1.00]	0.77 [0.53-1.00]	0.64 [0.33-0.95]	0.70 [0.43-0.96]	0.05 [0.00-0.51]	0.19 [0.00-0.65]
5	0.75 [0.52-0.97]	0.87 [0.75-0.99]	0.83 [0.67-0.99]	0.79 [0.60-0.98]	0.72 [0.47-0.97]	0.12 [0.00-0.55]	0.25 [0.00-0.70]
6	0.79 [0.59-0.98]	0.83 [0.66-1.00]	0.88 [0.76-1.00]	0.71 [0.42-1.00]	0.75 [0.52-0.98]	0.23 [0.00-0.69]	0.42 [0.00-0.90]
7	0.84 [0.69-0.99]	0.86 [0.72-1.00]	0.83 [0.67-1.00]	0.68 [0.41-0.95]	0.56 [0.17-0.95]	0.13 [0.00-0.59]	0.34 [0.00-0.76]
8	0.74 [0.51-0.97]	0.81 [0.62-1.00]	0.78 [0.56-1.00]	0.69 [0.43-0.95]	0.78 [0.58-0.99]	0.13 [0.00-0.63]	0.25 [0.00-0.75]
9	0.78 [0.57-0.98]	0.86 [0.73-1.00]	0.84 [0.68-0.99]	0.68 [0.40-0.96]	0.76 [0.52-0.99]	0.30 [0.00-0.76]	0.51 [0.14-0.88]
10	0.76 [0.53-0.98]	0.85 [0.72-0.99]	0.80 [0.62-0.98]	0.62 [0.30-0.94]	0.63 [0.29-0.97]	0.30 [0.00-0.77]	0.34 [0.00-0.81]
1-10	0.76 [0.56-0.96]	0.86 [0.74-0.98]	0.83 [0.69-0.97]	0.68 [0.42-0.93]	0.68 [0.43-0.92]	0.07 [0.00-0.37]	0.28 [0.00-0.64]
2-10	0.77 [0.57-0.96]	0.86 [0.74-0.98]	0.83 [0.69-0.97]	0.68 [0.42-0.93]	0.71 [0.47-0.94]	0.11 [0.00-0.43]	0.28 [0.00-0.67]

Table 2- Reliability of Stiffness Measures (ICC) for Trial, Session and Week

Table 3: Standard Error of Measurement between Weeks 1 and 2

Variable	Standard Error of Measure (SEM)	Minimal Detectable Difference (MDD)
Lower limb stiffness (kNm ⁻¹)	0.42	1.16
Ankle angle (degrees) 80 ms pre-contact (dorsiflexion)	0.97	2.69
Ankle angle (degrees) @ contact (dorsiflexion)	1.25	3.46
Ankle angle (degrees) @ take-off (dorsiflexion)	1.28	3.55
Ankle stretch amplitude (degrees)	1.45	4.02
Knee stretch amplitude (degrees)	5.39	14.94
Hip stretch amplitude (degrees)	1.26	3.49

Table 3- Standard Error of Measurement between Weeks 1 and 2

Illustrations

Figure 1: 3D-Motion Analysis Marker Set Configuration



Figure 2: Custom-Built Low-Friction Sledge Jump System

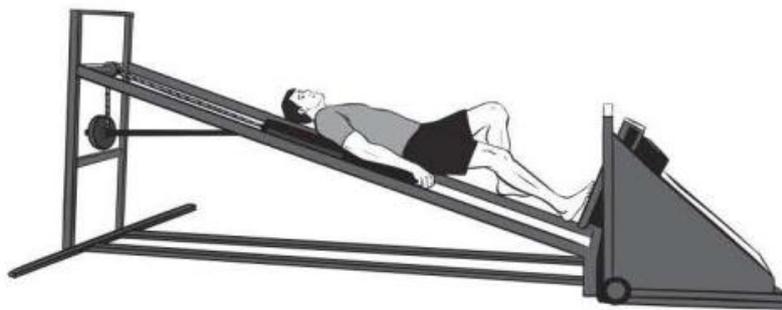


Figure 3: Lower Limb Stiffness Derivation

$$K_{\text{lowerlimb}} = \frac{M \times \Pi (t_a + t_c)}{(t_c^2 ((t_a + t_c / \Pi) - (t_c / 4)))}$$

Unit- kN.m^{-1}

M = total body mass
 t_a = flight time
 t_c = ground contact time

Figure 4: Temporal Reliability of Lower Limb Stiffness

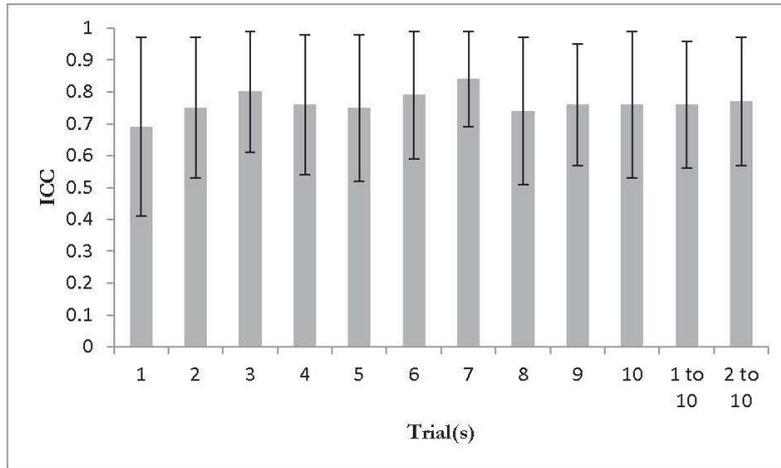


Figure 5: Temporal Reliability of Ankle Angles (80 ms Pre-Contact, Contact and Take-Off)

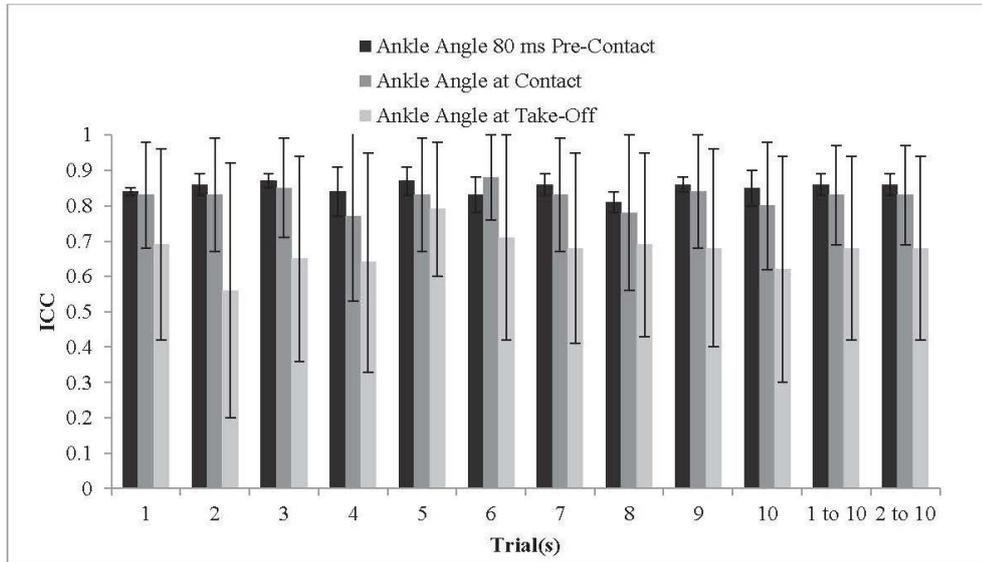
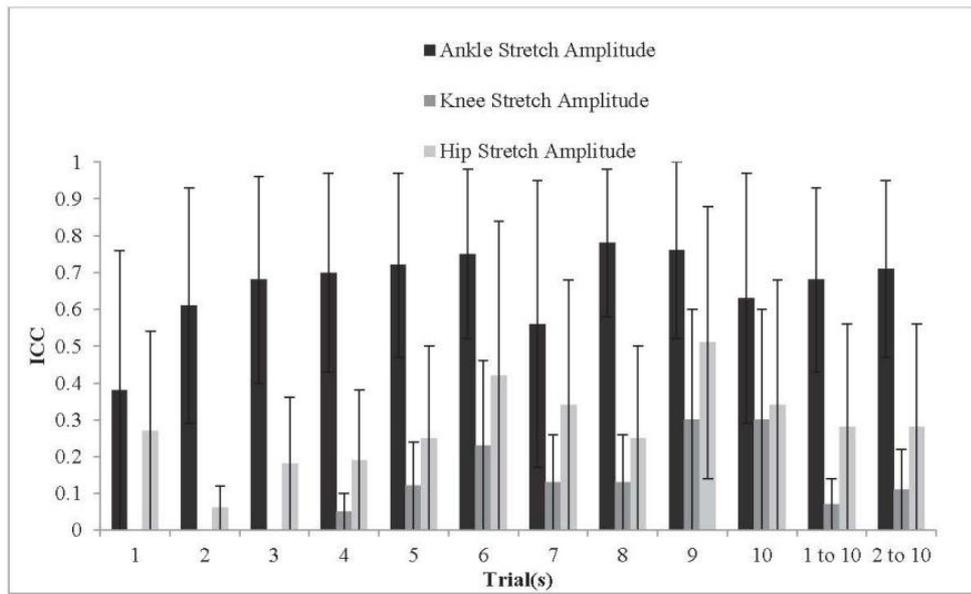


Figure 6: Temporal Reliability of Lower Limb Stretch Amplitude Measures (Ankle, Hip and Knee)



PARTICIPANT INFORMATION SHEET

Study Title:

Part 1- “An investigation into the reliability of measuring muscle and joint behaviour during single-limb hopping”

Part 2- “Are measures of ankle stiffness reliable between upright hopping and inclined sleigh hopping”

Principal investigators: James Debenham and Mervyn Travers

Co-investigators: Associate Professor Garry Allison; Dr William Gibson; Dr Amity Campbell

Purpose of Research

This study is part of a series of studies looking at the way in which muscles and joints work and behave when humans hop. The aim of the research is to help increase understanding of how some people sustain chronic lower limb injuries like Achilles tendon problems. It may also assist in the development of new strategies to prevent and/or treat this condition. The studies employ a device, known as an inclined-sleigh, or ‘sleigh’ for short. The sleigh allows subjects to freely hop on an incline with their bodyweight supported. This enables measurement of how the muscles and joints behave whilst subjects hop in a much more accurate way than during upright hopping. Because the sleigh has never been used before, the accuracy and reliability of the measurements being taken needs to be determined. By knowing this future studies can be performed in the knowledge of how precise the sleigh is for measuring muscle activity and joint movement.

Your Role

Testing takes place in the Motion Analysis Laboratory at Curtin University of Technology’s Bentley campus. You will be required to attend testing on 2 different occasions (Day 1 and 8). On each occasion you will be required to wear shorts and a t-shirt. You will have various markers placed on your legs to allow us to take measurements. You will be given an explanation and a demonstration regarding the hopping tasks you will be performing. You may ask questions at any point. You will be required to perform single-limb hopping for 10 sets of 30 seconds with a 90 second rest between each set. Following that you will be positioned on the sleigh and required to repeat the same task on the opposite leg. This process will be repeated identically on the subsequent occasion that we test. The whole process will take less than 90 minutes on each occasion. You will be requested to fill in a diary in the week between testing days to report any muscle soreness that may be associated with testing (see below Risks and Discomfort).

Risks and Discomfort

The various markers placed on your skin may cause some mild discomfort when they are removed. There is no inherent risk in the activity you will be requested to perform. You may feel some very mild fatigue towards the end of testing. The test is not designed to create any discomfort, although you may experience some muscle soreness if this type of exercise is novel for you. If this occurs, it will resolve within a few days.

Benefits

There is no direct benefit to you from participating in this study. However information gained from this study may ultimately provide insights into the mechanisms underlying certain musculoskeletal lower limb conditions. It may also aid in the development of rehabilitation protocols for such conditions in the general community.

Consent to Participate

Your involvement in this research is entirely voluntary. If you choose not to participate or at any point in the study you wish to withdraw, you have the right to do so without explanation, justification or consequence of any kind. After reading this form and asking any questions you may have, you will be asked to sign a consent form indicating you voluntarily agree to participate in this study and you are aware of all your rights in relation to participation.

Confidentiality

Your privacy is greatly respected. We will not ask any personal information of you beyond information such as height, weight, age and past musculoskeletal injury history. The data collected from you will be kept separate from your personal details, and only the investigators will have access to this. The data collected from you will have all identifying information removed. This is in adherence to University policy. All data will be securely stored for a period of 7 years, following which it will be destroyed. Any publications or presentations resulting from this study will be presented as averages, thus not personally identifiable to any one individual.

Further Information

This study has been approved by the Curtin University Human Research Ethics Committee (Approval Number PT0151/2009 (Part 1) and PT0145/2009 (Part 2)). The Committee is comprised of members of the public, academics, lawyers, doctors and pastoral carers. Its main role is to protect participants. If needed, verification of approval can be obtained either by writing to the Curtin University Human Research Ethics Committee, c/- Office of Research and Development, Curtin University of Technology, GPO Box U1987, Perth, 6845 or by telephoning 9266 2784 or by emailing hrec@curtin.edu.au . If you would like further information about the study, please feel free to contact me on 08 9266 3667 or by email: j.debenham@curtin.edu.au, m.travers@curtin.edu.au

Thank you very much for your involvement in this research. Your participation is greatly appreciated.

Consent Form

This study has been approved by the Curtin University Human Research Ethics Committee (Approval Number PT0151/2009 (Part 1) and PT0145/2009 (Part 2)).

- I understand the purpose and procedures of the study
 - I have been provided with the participation information sheet
 - I understand that the study itself may not benefit me
 - I understand that my involvement is voluntary and I can withdraw at any time without problem
 - I understand that no personal identifying information like my name and address will be used and that all information will be securely stored for 7 years before being destroyed
 - I agree that research gathered from this study may be published provided that any information that may identify me is not used
 - I have been given the opportunity to ask questions
 - I agree to participate in the study outlined to me
-

Name: _____ Signature: _____ Date: _____

Witness: _____ Signature: _____ Date: _____

Participant Information Document

Study Title: Joint position sense during the stretch-shortening cycle

Project investigator Mervyn Travers – PhD Candidate, Curtin University of Technology

Supervisors Associate Professor Garry Allison
Dr. Will Gibson
Dr. Amity Campbell

Purpose of Research

The purpose of this project is to investigate how sensitively people perceive changes in landing surface height during inclined sleigh hopping.

Your Role

Time requirements

You will be asked to attend the School of Physiotherapy at Curtin University of Technology on one test occasion. Testing is expected to take 90 minutes.

Testing protocol

The following protocol will be followed:

Firstly, you will be requested to sign a consent form.

Reflective markers will be applied using tape to your pelvis and legs to allow 3-D motion analysis. This is a commonly used, comfortable and safe process.

You will be requested to perform multiple trials of 5 hops on a force plate attached to an inclined sleigh apparatus under 2 hopping conditions – alternate & bi-lateral. A random, small surface height change will be introduced when you are hopping. You will be informed on which leg and in which direction (up or down) the changes will occur for each trial of 5 hops. You will be requested to inform the researcher when you perceive a change in surface height. This will be repeated until the examiner has determined the minimal detectable difference for each hopping condition 3 times

You will be blind-folded and wearing headphones playing music to eliminate visual and auditory feedback regarding surface changes.

You will be given a 15 second rest period between each trial.

Risks and Discomfort:

You may experience mild fatigue in the calf muscles as a result of the testing. You are not expected to experience any pain or discomfort associated with testing.

Benefits

The information collected in this study has the potential to assist in increasing health professionals understanding of how the motor system of the lower limb operates.

Confidentiality

All study information will be stored by the project supervisor, Assoc. Prof. Garry Allison, in a secure place for 5 years at the School of Physiotherapy. After this period all the transcripts will be destroyed, this is in accordance with requirements of Curtin University of Technology.

Consent to Participate:

Your participation in this research project is completely voluntary and you may withdraw at any time without prejudice or negative consequences. If you do wish to withdraw please contact the study investigator at your earliest opportunity at Tel: 9266 3667.

Further Information

This study has been approved by the Curtin University Human Research Ethics Committee (PT0145). The Committee is comprised of members of the public, academics, lawyers, doctors and pastoral carers. Its main role is to protect participants. If needed, verification of approval can be obtained either by writing to the Curtin University Human Research Ethics Committee, c/- Office of Research and Development, Curtin University of Technology, GPO Box U1987, Perth, 6845 or by telephoning 9266 2784 or by emailing hrec@curtin.edu.au. If you would like further information regarding the study, feel free to contact me at 08 9266 3667 or m.travers@curtin.edu.au

Thank you very much for your involvement in this research. Your participation is greatly appreciated.

Consent Form

Study Title: Joint Position Sense during the stretch shortening cycle

Project investigator Mervyn Travers – PhD Candidate, Curtin University of Technology

Supervisors Associate Professor Garry Allison
Dr. Will Gibson
Dr. Amity Campbell

I, _____
(PLEASE PRINT)
consent to my participation in this study.

I have read the Participant Information Document and understand the consequences and risks associated with involvement in this study.

I understand the purpose and procedure of this study.

I understand that the study may not benefit me directly.

I have had the opportunity to ask and have answered to my satisfaction any questions regarding all aspects of this study.

I give permission for any results from this study to be used in any report or research paper, on the understanding that my anonymity will be preserved.

I understand I retain the right to withdraw from this study at any time, and without prejudice.

I understand that the questionnaire data I provide during this study will be confidential and will not be given to my employer or clients.

Version 1
10/11/09

1 of 2

Signature _____

Date _____

Witness _____

Date _____

This study has been approved by the Curtin University Human Research Ethics Committee (Approval Number PT0145).

The Committee is comprised of members of the public, academics, lawyers, doctors and pastoral carers. Its main role is to protect participants. If needed, verification of approval can be obtained either by writing to the Curtin University Human Research Ethics Committee, c/- Office of Research and Development, Curtin University of Technology, GPO Box U1987, Perth, 6845 or by telephoning 9266 2784 or by emailing hrec@curtin.edu.au

All participants are informed that, if they have any complaint regarding the manner, in which a research project is conducted, it may be given to the research supervisor (Assoc Prof GT Allison 9266 3626) or, alternatively to the Secretary, Human Research Ethics Committee, School of Physiotherapy, Curtin University (telephone number 92662784). All study participants will be provided with a copy of the Information Sheet and Consent Form for their personal records.

Version 1
10/11/09

2 of 2

PARTICIPANT INFORMATION SHEET

Study Title: The role of co-contraction as a mechanism for modulating ankle joint stiffness

Project investigator Mervyn Travers
Supervisors Professor Garry Allison
Dr. Will Gibson
Dr. Amity Campbell

Purpose of Research

This study aims to investigate how the ankle reacts to varying hopping tasks. We are particularly interested in how the muscles react to small changes in the floor surface whilst hopping. This research may help us to better understand movement and may allow for the improvement of rehabilitation protocols in the future.

Your role

You will be asked to attend the School of Physiotherapy at Curtin University of Technology on one test occasion. Testing is expected to take approximately 2 hours. You will be requested to wear shorts and a t shirt for testing.

The following protocol will be followed:

Firstly, you will be required to sign a consent form.

Your skin shall be prepared for surface electromyography – this entails shaving, exfoliating, alcohol swabbing and applying surface electrodes to three small areas of skin on your leg – on the front of your shin, your calf and ankle. This is a gentle and pain free process which ensures measures of muscle activity and timing of your lower leg muscles can be measured.

Reflective markers will be applied using tape to your hips, legs and feet to allow 3-D motion analysis. This is a commonly used, comfortable and safe process. You will then be given a 10 minute period to practice, ask questions and become familiar with the testing procedures.

You will be requested to perform 4 trials of 10 successive bilateral hops on an inclined sleigh. Whilst hopping you will be asked to keep your eyes closed. You will also be wearing headphones playing music. 4 hopping conditions will be randomly used:

1. Hopping on both feet at your comfortable pace
2. Hopping on both feet spending as little time on the ground as possible
3. Hopping on both feet at your comfortable pace, during this trial one 6mm increase in floor height will occur under the dominant foot
4. Hopping on both feet at your comfortable pace, during this trial one 36mm increase in floor height will occur under the dominant foot

You will have a 2 minute rest between each trial. When you have completed all 4 trials the test will be finished.

Risks and Discomfort

You are not expected to experience any discomfort during or as a result of testing.

Benefits

The information collected in this study has the potential to assist in increasing health professionals understanding how the ankle reacts to changes in surface height. This has application from daily activities to sporting activities.

Confidentiality

Only the researchers listed above will have access to the data from this study. The data will be de-identified so that the participants cannot be identified. The data is intended for the preparation of scientific papers for publication in scholarly journals. All study information will be stored by the project supervisor, Prof. Garry Allison, in a secure place for 5 years at the School of Physiotherapy. After this period all the transcripts will be destroyed, this is in accordance with requirements of Curtin University of Technology.

Refusal or Withdrawal:

Your participation in this research project is completely voluntary and you may withdraw at any time without prejudice or negative consequences. If you do wish to withdraw please contact the study investigator at your earliest opportunity on Tel 92663667 or email m.travers@curtin.edu.au

Further Information

This study has been approved by the Curtin University Human Research Ethics Committee (HR 61/2011). The Committee is comprised of members of the public, academics, lawyers, doctors and pastoral carers. Its main role is to protect participants. If needed, verification of approval can be obtained either by writing to the Curtin University Human Research Ethics Committee, c/- Office of

Research and Development, Curtin University of Technology, GPO Box U1987, Perth, 6845 or by telephoning 9266 2784 or by emailing hrec@curtin.edu.au

Informed Consent Document

Study Title: The role of co-contraction as a mechanism for modulating ankle joint stiffness

Project investigator: Mervyn Travers
Supervisors: Professor Garry Allison
Dr. Will Gibson
Dr. Amity Campbell

I, _____,

(PLEASE PRINT)

consent to my participation in this study. I have read the Participant Information Sheet and understand the consequences and risks associated with involvement in this study. I have had the opportunity to ask and have answered to my satisfaction any questions regarding all aspects of this study.

I give permission for any results from this study to be used in any report or research paper, on the understanding that my anonymity will be preserved. I understand I retain the right to withdraw from this study at any time, and without prejudice.

SIGNED _____ Date _____

This study has been approved by the Curtin University Human Research Ethics Committee (Approval Number 61/2011).

The Committee is comprised of members of the public, academics, lawyers, doctors and pastoral carers. Its main role is to protect participants. If needed, verification of approval can be obtained either by writing to the Curtin University Human Research Ethics Committee, c/- Office of Research and Development, Curtin University of Technology, GPO Box U1987, Perth, 6845 or by telephoning 9266 2784 or by emailing hrec@curtin.edu.au

All participants are informed that, if they have any complaint regarding the manner, in which a research project is conducted, it may be given to the research supervisor (Prof GT Allison 9266 3626) or, alternatively to the Secretary, Human Research Ethics Committee, School of Physiotherapy, Curtin University (telephone number 92662784). All study participants will be provided with a copy of the Information Sheet and Consent Form for their personal records.

Appendix 5

IF PROPRIOCEPTION IS RELEVANT TO LOCOMOTION, THEN WHY TEST IT STANDING STILL?

Mervyn Travers¹, James Debenham², Dr. William Gibson², Dr. Amity Campbell¹ and Prof. Garry Allison¹

¹*School of Physiotherapy and Exercise Science, Curtin University, GPO box 1987, WA 6845, Australia*

²*University of Notre Dame, School of Physiotherapy, 19 Mouat Street, Fremantle, WA 6959, Australia*
m.travers@curtin.edu.au, james.debenham@nd.edu.au, william.gibson@nd.edu.au, a.campbell@curtin.edu.au,
g.allison@curtin.edu.au

Keywords: Proprioception, stretch-shortening cycle, minimal perceptible difference

Abstract: Traditionally, proprioception research has utilised passive position or movement detection and repositioning tasks. Current evidence suggests proprioception represents a complex synergy of sensory inputs that may be more appropriately assessed during more functional tasks. This study investigated the Minimal Perceptible Difference (MPD) test - a novel assessment of participants' ability to perceive floor height changes whilst hopping. Sixteen healthy volunteers performed multiple hopping trials on a custom-built sleigh apparatus that permitted a floor height change (range 3mm to 48mm). The MPD in floor height was recorded for 8 different hopping conditions (Factors - Technique: alternate / bilateral hopping; Side: dominant / non dominant; Direction of change: up / down) over two separate testing occasions. Within and between-day reliability were assessed using ICC and 95% confidence intervals. Hopping technique was the only factor which significantly influenced participants' sensitivity to detect changes in floor height. The mean MPD was significantly lower ($p < .0001$) for bilateral hopping (15.65mm) when compared to alternate hopping (26.59mm). Bilateral hopping yielded strong ICC for within and between day reliability. We propose the bilateral hopping MPD assessment is a reliable, functional assessment of proprioception sensitivity that may better reflect human gait than established static assessments.

1 INTRODUCTION

Proprioception is defined as "afferent information from proprioceptors located in the proprioceptive fields that contribute to conscious sensations (muscle sense), total posture (postural equilibrium) and segmental posture (joint stability)" (Sherrington, 1906). It is a concept with particular relevance to clinicians and researchers with respect to performance, injury and rehabilitation (Brumagne et al., 2004, Cameron et al., 2008, Fu and Hui-Chan, 2007, Herrington et al., 2008, Lephart and Jari, 2002, Vuillerme and Boisgontier, 2008). Most methodologies examine proprioceptive acuity using passive position detection (Down et al., 2007), passive movement detection (Salles et al., 2011) and active repositioning tasks (Ribeiro et al., 2007). However, recent evidence suggests proprioception is a much more complex concept than just joint position sense and kinaesthesia, incorporating the

integration of the body schema (Ivanenko et al., 2011) and its continuous refinement being expanded to a concept called "somatoperception" (Longo et al., 2010). Of particular note is that current testing methods may represent convenient research methods, but may not reflect the dynamic function of the lower limb - to perform repeated Stretch Shortening Cycles (SSC) (Proske et al., 2000).

The authors utilised a sub-maximal sleigh hopping model to replicate the normal function of the limbs via repeated SSC. This model has been applied to develop the Minimal Perceptible Difference (MPD) test - a novel research tool which examines individuals' ability to detect changes in floor surface height during the repeated SSC. This study aimed to investigate the reliability of the MPD test on a within and between day basis. We also aimed to quantify the MPD in floor surface height for a healthy population.

2 METHODS

The MPD test examined the sensitivity of healthy participants to perceive changes in floor surface height whilst hopping on a custom built apparatus (Figure 1) with an adjustable floor (Figure 2).

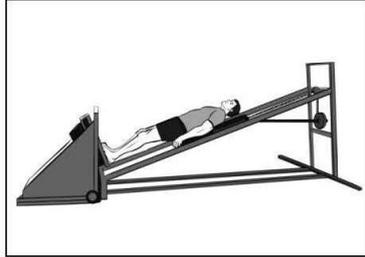


Figure 1: The Custom Built Sleigh Apparatus



Figure 2: The Sliding Floor Mechanism illustrating a 36mm reduction in floor height under the right foot

Sixteen healthy participants performed multiple trials of 5 consecutive hops on a custom built sleigh apparatus that permitted the testers to change the floor height (range 3mm to 48mm) during each trial, as dictated by a structured searching algorithm.

MPD in floor height was recorded for 8 different hopping conditions (Factors - Technique: alternate / bilateral hopping; Side: dominant / non dominant; Direction of change: up / down) over two testing occasions spaced one week apart. Participants performed a mean of 117 trials on Day 1 and 120 trials on Day2.

3 RESULTS

Intra-class Correlation Coefficients (ICC) and 95%

Confidence Intervals (CI) were derived to examine within and between day reliability of the MPD test. All bilateral hopping techniques yielded moderate to high ICC values for both within (0.60 to 0.79) and between day (0.67 to 0.88) reliability.

The only factor which significantly influenced the sensitivity of subjects to detect changes in floor height was the hopping technique (bilateral or alternate, $p < 0.05$). Comparing across hopping techniques, the mean MPD was significantly lower ($p < .001$) for bilateral hopping than alternate hopping as per table 1:

Table 1: Result of Linear Mixed Model Analysis of Between Days MPD Scores (in mm).

MPDmean Bilateral Hopping	MPDmean Alternate Hopping	Significance
15.65	26.59	$p < .001$

4 DISCUSSION

We propose the MPD test is a novel, reliable and functionally relevant research tool. Furthermore, the MPD represents a change in the research paradigm from testing detection of passive, position matching and force matching of isolated joints (Down et al., 2007, Jong et al., 2005, Lowrey et al., 2010, Matre et al., 2002). Instead we examined proprioception during repeated SSC which may better represent human gait as it considers an expected interaction with a non-homogenous interface between the foot and ground.

For bilateral hopping, ICC values for within and between day comparisons all exceeded proposed ICC of 0.6 that has been recommended for any measure to have clinical utility (Chinn, 1991). This indicates that the MPD test using bilateral hopping may have application in the research setting on single and multiple test occasions.

The increased sensitivity to floor height change detection during bilateral hopping is an interesting observation. Our findings are consistent with previous findings which suggest that gating and utility of sensory information may be strategy dependant (Ivanenko et al., 2000).

It may be hypothesised that the bilateral tasks represent upright stance where we need sensitive and constant feedback to maintain posture and to safely initiate movement. Continuous weight bearing feedback during bipedal stance may be provided via continuous bilateral comparison of both limbs contributing to the postural schema.

Conversely, reduced sensitivity to floor height change detection during alternate hopping suggest that bipedal gait may allow humans progress their centre of mass with sufficient proprioceptive redundancy to overcome large variations in the interface between the foot and the ground surface without cognitive perception of the challenge.

5 FUTURE RESEARCH

We have observed a significant difference in sensitivity to detect floor height changes between alternating and bilateral SSC. An area for future research is to investigate whether this difference in detection is attributable to neurological or biomechanical factors. Furthermore, given these observations more research is required to determine if current “static tests” are valid correlates to dynamic activities such as gait.

6 CONCLUSIONS

The MPD test has been presented and represents a change in research focus towards investigating proprioception using repeated stretch shortening cycle to model normal lower limb dynamic function.

Development of this tool may allow for further investigation of functional proprioceptive ability injured/pathological samples. The MPD test has been demonstrated as reliable over time and is therefore an acceptable research tool for use within and across test occasions.

We observed greater sensitivity of the MPD test in the bilateral hopping technique. This may reflect specific sensory requirements for upright stance, whereas (bipedal) gait may have its own specific redundancies.

ACKNOWLEDGEMENTS

This research was partly supported by a research grant from the Neurotrauma Research Program – WA.

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Appendix 6

CONSCIOUS DRIVE TO STIFFEN THE LEG SPRING – MOTOR STRATEGIES FOR AN INTERNAL CHALLENGE

Mervyn Travers¹, James Debenham², Dr. William Gibson², Dr. Amity Campbell¹ and Prof. Garry Allison¹

¹*School of Physiotherapy and Exercise Science, Curtin University, GPO box 1987, WA 6845, Australia*

²*University of Notre Dame, School of Physiotherapy, 19 Mouat Street, Fremantle, WA 6959, Australia*
m.travers@curtin.edu.au, james.debenham@nd.edu.au, william.gibson@nd.edu.au, a.campbell@curtin.edu.au,
g.allison@curtin.edu.au

Keywords: Feed-forward, stretch-shortening cycle, stiffness, internal challenge

Abstract: This study investigated the kinematic and muscle activity profiles at the ankle under two hopping conditions that consciously altered leg stiffness. Nine healthy volunteers performed multiple trials of bilateral hopping on a custom built sleigh under two conditions – preferred (PC) and short contact (SC). Leg stiffness, peak EMG, time to peak EMG and co-activation ratios for the medial gastrocnemius (MG), soleus (Sol) and tibialis anterior (TibAnt) muscles were compared across conditions. SC hopping resulted in increased leg stiffness. Importantly, Sol onset shifted from 86ms post-contact during PC to 14ms post-contact for SC. Similarly, MG onset was 41ms post-contact during PC and 22ms pre-contact for SC. Significantly earlier onsets of Sol and MG represent a shift into the feed-forward window which was not reflected by TibAnt. Comparisons revealed no significant differences in co-activation ratios ($p > 0.05$) suggesting that increased leg stiffness during SC hopping was not a result of increased co-activation. Instead a dynamic strategy pairing pre-activation with an increased rate of activity of the agonist muscles to develop force in time for contact with the surface was observed. We suggest that the optimal strategy to consciously drive increased leg stiffness occurs via a feedforward response.

1 OBJECTIVES

It is well-established that simultaneous contraction of primary agonist and antagonist muscle groups (i.e. muscle co-activation) will increase the stiffness of a joint (Blickhan, 1989). This process is considered protective, for example when landing (Santello, 2005, Yeadon et al., 2010). Furthermore, such co-activation has been observed at the knee in athletes following anterior cruciate ligament repair (Bryant et al., 2009) and in the lumbar region in clinical pain cohorts (Hodges et al., 2009, Morris et al., In Press: 2013, Moseley et al., 2004, van Dieën et al., 2003). The common experimental paradigm for investigating stiffness modulation utilises external challenges including hopping and running on surfaces of varying rigidity (Ferris et al., 1999, Ferris and Farley, 1997, Moritz et al., 2004), running on uneven surfaces (Müller and Blickhan, 2010) and even reduced cutaneous feedback during hopping (Fiolkowski et al., 2005).

The existing literature does not consider the potential differences in motor strategies responsible for stiffness modulation in response to internal challenges. It cannot be assumed that humans would utilise the same motor strategies to adapt to both internal and external challenges. In fact, a previous study has used conscious effort (an internal challenge) to increase stiffness during hopping by reducing ground contact time (Hobara et al., 2007). Interestingly, they observed increased stiffness without muscle co-activation (Hobara et al., 2007) suggesting that the conscious drive to produce a “stiffer” performance may have its own unique motor strategies. Yet, these specific motor strategies remain unstudied despite being relevant to optimising running performance (Hobara et al., 2010). The specific motor patterns responsible for stiffness modulation under internal challenges are relevant to performance, injury and rehabilitation.

The ankle joint is the major determinant of lower limb stiffness during low load tasks (Farley and Morgenroth, 1999, Moritz et al., 2004). This study

examined the muscle activity profile changes at the ankle associated with consciously driven increase in leg stiffness during repeated submaximal hopping.

2 METHODS

This study utilised a within-subject experimental design. Nine healthy participants performed multiple hopping trials on a Custom Built Sleigh Apparatus (Figure 1). The sleigh incorporated an instrumented (AMTI forceplate 1kHz sampling) landing platform allowing the establishment of event markers.

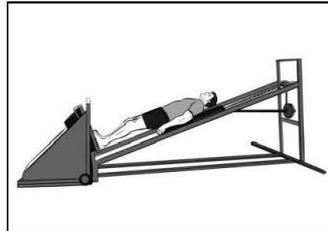


Figure 1: Double leg hopping on the Custom Built Sleigh Apparatus inclined at 20degrees from horizontal.

Each Trial involved 10 continuous bilateral hops on the sleigh apparatus (median 6 used for analysis). Participants minimised any associated knee flexion (no external fixation was used) so that performance was primarily driven by the ankles. Three trials were performed under two different conditions – preferred ground contact time (PC), and with as short a contact time as possible (SC).

Surface EMG for the medial gastrocnemius (MG), soleus (SOL), and the tibialis anterior (TibAnt) muscles was collected using an AMT-8 (Bortec Biomedical Ltd.) system. The EMG signal was full wave rectified and onsets detected using the integrated protocol (Allison, 2003). Trial linear envelopes were created using a fourth-order zero-lag, Butterworth low-pass filter (10 Hz) and temporally synchronised to foot contact. Ensemble average LE were determined for a 760ms window (280ms pre-contact to 480ms post-contact). The feedforward window was defined at 33ms post-contact (Voigt et al., 1998). EMG signals were integrated in 20ms epochs (IEMG) for the 760ms window. The median peak of 10 PC hops was used as a 1.0 arbitrary unit for amplitude normalisation (Allison et al., 1993).

Leg Stiffness (K) was calculated using the formula below (Dalleau et al., 2004):

$$K = \frac{M \times \Pi(t_f + t_c)}{(t_c^2 ((t_f + t_c / \Pi) - (t_c / 4)))}$$

Figure 2: Formula for estimating leg stiffness (K); M = body mass; t_f = flight time; t_c = ground contact time

Co-activation was defined as the ratio of the agonist (MG and Sol) and antagonist (TibAnt) muscle activity and co-activation ratios were labelled MG/TibAnt and Sol/TibAnt.

Paired samples t-tests were used to compare differences in K, co-activation ratios and onset times between conditions. A linear mixed model was utilised to identify any significant difference in onset times for each muscle grouped for condition and side. It was further used to investigate any interaction between condition, side and muscle with onset time as the dependant variable.

3 RESULTS

The participants demonstrated a significant increase in K during SC hopping (Table 1).

Table 1: Leg Stiffness (kN.m-1) by Condition

Condition	Stiffness (SD)	p
PC	9.20 (2.58)	
SC	14.16 (3.09)*	p < .001

The peak EMG amplitude was not significantly different for any muscle between PC and SC. Further, there was no significant interaction between side and hopping condition ($F = .182$, $p = .671$) nor main effects for side ($F = .284$, $p = .596$) or condition ($F = .690$, $p = .409$).

Sides were pooled for EMG onsets and time to peak EMG as there was no significant interaction between side and muscle or condition ($p > 0.05$).

The Sol onset time was 86ms (95% CI 58ms to 114ms) post-contact for the PC condition and 14ms (95% CI -7ms to 36ms) post-contact for the SC condition ($F = 58.145$, $p < .001$). The MG onset time was 41ms (95% CI 25ms to 57ms) post-contact for the PC condition and 22ms (95% CI 35ms to 9ms) post-contact for the SC condition ($F = 56.137$, $p < .001$). The TibAnt onset was not altered significantly between conditions ($p = .062$).

Peak Sol activity occurred at 200ms (95% CI 184ms to 216ms) post-contact for PC and was significantly earlier ($p < .001$) for SC occurring at 114ms post-contact (95% CI 101ms to 125ms). Similarly, peak MG activity occurred at 195ms post-

contact (95% CI 179ms to 211ms) and was significantly earlier ($p < .001$) for SC occurring at 102ms post-contact (95% CI 91ms to 114ms). The time to peak activity for TibAnt was not significantly different between conditions ($p < .05$).

Finally, comparisons revealed no significant differences in MG/TibAnt and Sol/TibAnt co-activation ratios between PC and SC ($p > 0.05$).

4 DISCUSSION

Both MG and Sol demonstrated earlier onsets during SC hopping. This represented a change from potential feedback latency to a clear feedforward response with onsets occurring within the defined 33 (+/- 7)ms window (Voigt et al., 1998). Importantly, this was not matched by TibAnt.

Specifically, our findings demonstrate that in the presence of a controlled environment and self-regulation of the pending challenge and consequences (i.e. the choice of hopping contact time on a stable sleigh) individuals may choose a feedforward strategy instead of the established co-activation strategy. We observed a dynamic strategy of pre-activation with an increased rate of activity of the agonist muscle to develop force in time for contact with the surface.

5 CONCLUSIONS

This study investigated the neural control of consciously driven increase in joint stiffness during submaximal hopping. We observed a stiffer hopping performance driven by a feedforward strategy confirming our hypothesis that internal challenges to performance have their own unique motor strategies.

ACKNOWLEDGEMENTS

This research was partly supported by a grant from the Neurotrauma Research Program – WA.

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