VALIDITY AND RELIABILITY OF INERTIAL MEASUREMENT UNITS IN OBTAINING A LOWER LIMB STIFFNESS RUNNING MEASURE IN HIGH-LEVEL TRACK AND FIELD ATHLETES

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STATEMENT OF SOURCES

This thesis contains no material published elsewhere in whole or in part from a thesis by which I have qualified for or have been awarded in another degree.

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

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All research procedures in this thesis received the approval from the Australian Catholic University Research Ethics Committee (Approval no. 2016-284H).

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ABSTRACT

Introduction: Recent developments in inertial sensor technology mean real time monitoring and tracking of athletes in the daily training environment is now a possibility. Such developments have the potential for injury prevention and performance maximisation. Stiffness of the lower limb has known links to performance and injury risk; however, these measures have so far been limited to laboratory-based settings. Application of current sensor technology has the potential for ongoing stiffness assessment not only in the laboratory but also in the daily training environment. Actual training monitoring and changes to the way an athlete deals with loading (leg stiffness) on a regular basis could provide vital feedback to athletes, coaches, medical and support staff allowing for effective systems to be put in place to ensure athletes reach their potential.

Study 1: The first aim of this thesis was to review existing literature surrounding the longitudinal assessment of lower limb stiffness in adult athletic populations. A systematic review was conducted which initially produced 630 results before being reduced to 6 for final analysis, highlighting the lack of research in this area. Data extracted focused on the population, methodologies and key findings of each study. The results concluded that the longitudinal assessment of lower limb stiffness had so far been isolated to laboratory-based settings and predominately measured through simple vertical hopping and jump tasks in the specific sporting population of Australian Rules Football players. From the results, the need for a field-based measure of lower limb stiffness was identified in order for stiffness to be assessed at more regular intervals to better understand the prospective links between lower limb stiffness, performance and injury.

Study 2: Based on the findings from study 1, the primary aim of this study was to develop a valid and reliable field-based measure of lower limb stiffness in high-level track and field athletes during running (a task reflective of training and competition) using inertial measurement units. Nineteen high-level track and field athletes completed six running gait trials at a pace reflective of their event during competition. Data was captured using a fourteen-camera motion analysis system (250Hz), a force plate (1000Hz) and three inertial measurement units (500Hz). The gold standard stiffness measures from the motion analysis system were than compared with the stiffness measures derived from the inertial measurement units. Poor validity was found between the gold standard stiffness measures and the measures derived from the inertial measurement units. In addition, the results demonstrated that the data output from the inertial measurement units were not reliable when substituted into the existing measures of stiffness, warranting the need for further research.

Conclusion: This thesis makes a novel contribution to the assessment of lower limb stiffness in athletic populations. Although a valid and reliable measure of lower limb stiffness using inertial measurement units still needs to be established, it is hoped that this research is the first step in developing a daily monitoring tool which may provide a proactive approach in managing an athlete's response to load. However, further refinements to the algorithm and developments in inertial sensor technology are required before this technology can be considered for use outside the traditional laboratory setting.

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LIST OF DEFINITIONS

The following list comprises of the most commonly cited terms within this thesis and are defined relative to their use within the context of the topic.

High-level: An athlete who has competed at the national level or above, relative to age.

Inertial measurement unit/inertial sensor: A small electronic device consisting of an accelerometer, gyroscope and magnetometer that is fixed onto a human's limb segment to measure linear or angular motion.

Injury: A musculoskeletal problem that occurs during participation of an athlete's sport, ultimately leading to modified/missed training sessions or competition.

Lower limb stiffness/Leg Stiffness: A 'functional' form of stiffness that considers neuromuscular control and physical tissue properties. It is quantified as the peak force applied divided by the subsequent deformation of the 'spring' (i.e. leg). It is often used to calculate stiffness during horizontal movements, more specifically during running gait.

Sport-specific: A task that is reflective of performance demands of an athlete's training and competition environment.

Vertical Stiffness: A form of stiffness that aims to model the vertical displacement of the centre of mass. It is quantified as the peak vertical force divided by maximum vertical displacement. It is often used to calculate stiffness during direct vertical movements such as hopping and jumping.

LIST OF ABBREVIATIONS

3D	Three dimensional
AFL	Australian Football League
ANOVA	Analysis of variance
App	Application
COM	Centre of Mass
CV	Coefficient of variation
ES	Effect size
GPS	Global Positioning System
Hz	Hertz
ICC	Intraclass correlations
INJ	Injured
K_{joint}	Joint stiffness
K_{leg}	Leg stiffness
K_{vert}	Vertical stiffness
MTU	Muscle-tendon unit
NON-INJ	Non-injured
OR	Odds ratio
PSIS	Posterior superior iliac spine
PVGRF	Peak vertical ground reaction force
RR	Relative risk

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CHAPTER 1: INTRODUCTION

1.1 SYNOPSIS

Advancements in inertial sensor technology mean real time monitoring and tracking of athletes in the daily training environment is now a possibility. Such developments have the potential for performance enhancement and injury prevention. Lower limb stiffness has known links to performance and injury risk; however, these measures have been isolated to laboratory-based settings, which limit the scope of data collected and may influence natural running patterns (Little, Lee, James, & Davison, 2013). Application of current sensor technology has the potential for ongoing stiffness assessment not only in the laboratory but also the daily training environment.

Measures of lower limb stiffness view the leg as a simple spring mechanism, where a force is required to stretch or compress the spring (Butler, Crowell, & Davis, 2003). The compliance or resistance of this spring (i.e. leg) can be calculated using force and displacement measures providing a lower limb stiffness measure (Butler et al., 2003). Stiffness measures are particularly relevant for track and field performance as an increase in leg stiffness has been associated with an increase in jump height (Arampatzis, Schade, Walsh, & Bruggemann, 2001), running velocity (Kuitunen, Komi, & Kyröläinen, 2002), stride frequency (Farley & González, 1996) and running economy (Spurrs, Murphy, & Watsford, 2003).

Whilst increased lower limb stiffness improves performance (Arampatzis, Schade, et al., 2001; Avela & Komi, 1998; Farley & González, 1996; Kuitunen et al., 2002; Spurrs et al., 2003), too much or too little stiffness may influence an individual's risk of injury (Arampatzis, Brüggemann, & Klapsing, 2001; Butler et al., 2003; Williams, Davis, Scholz, Hamill, & Buchanan, 2004; Williams, McClay, & Hamill, 2001). However, the direct relationship between stiffness and injury remains somewhat unclear with higher levels of stiffness associated with bone related injuries while lower levels of stiffness have been linked with soft tissue injuries (Butler et al., 2003; Watsford et al., 2010).

Despite links between lower limb stiffness, performance and injury risk; longitudinal assessment of lower limb stiffness and its links with performance and injury remain scarce. In the few studies that prospectively assessed leg stiffness and injury, leg stiffness and limb asymmetry have been linked to potential injury risk in Australian Football League (AFL) players (Pruyn et al., 2012; Watsford et al., 2010). Although links between acute stiffness changes and performance exist, longitudinal changes to stiffness and performance remain unclear in athletic populations. Additionally, to date, measures of lower limb stiffness in athletic populations have come from simple vertical hopping and jump based tasks (Hobara et al., 2008; Pruyn et al., 2012, 2013; Serpell, Scarvell, Ball, & Smith, 2014; Watsford et al., 2010) which may not reflect lower limb stiffness during the daily training and competition environments.

The monitoring of training load and changes to the way an athlete manages loading (leg stiffness) could provide vital feedback to athletes, coaches, medical and support staff enabling early risk identification and performance maximisation. The information provided in real time to key high performance personnel could allow for effective systems to be put in place to ensure athletes reach their potential.

Inertial sensor technology allows acceleration, heading and orientation to be captured and data logged in a small unit (Hood, McBain, Portas, & Spears, 2012; Seel, Raisch, & Schauer, 2014). Accelerations have the potential to calculate a range of measures including step frequency and velocity (Hausswirth, Le Meur, Couturier, Bernard, & Brisswalter, 2009). There is an opportunity for stiffness measures to be derived from accelerometer sensor technology via contact/flight time and displacement force measures. The application of this novel technology has the potential for a proactive approach to the management of athlete performance and injury (i.e. reducing lost training time). Thus, the main objective of this thesis was to validate and implement a novel sensor-based measure to quantify lower limb stiffness in high-level track and field athletes.

1.2 AIM AND HYPOTHESIS

The primary aim of this research was to develop a valid and reliable measure of lower limb stiffness in high-level track and field athletes during running (a task reflective of training and competition) using inertial measurement units. It was hypothesised that measures of lower limb stiffness using inertial measurement units would be valid and reliable when compared to the gold standard stiffness measures derived from a three-dimensional (3D) motion analysis system and force plate.

1.3 SIGNIFICANCE

The use of inertial measurement units as a performance and injury monitoring tool has the ability to enhance service delivery and long-term development of athletes. It is hoped that this thesis will have a positive impact across multiple track and field-based disciplines with future potential to roll out across other running based sports (i.e. netball, basketball, soccer) to add vital information regarding an athlete's response to load allowing for early identification of injury risk. Whilst this thesis primarily focuses on the development of a valid and reliable field-based measure to quantify lower limb stiffness, significant potential outcomes of this research include:

- Daily monitoring of athletes
- Development of a tool for early identification of injuries
- Performance tracking (i.e. how an athlete responds to load)

In addition, the proposed measures may allow for remote monitoring for coaches, medical and service providers. The early detection and prevention of potential injury mechanisms may enhance athlete potential and career longevity.

1.4 LIMITATIONS

The following limitations are acknowledged:

1. Although the recruitment of high-level athletes may be a strength, the small sample size reduces the inferential capacity of outcomes.
2. The positioning of ground mounted force plates in the laboratory limited the amount of contacts captured by Vicon to one per trial.
3. The dropout of contacts from the IMeasureU sensors limited the amount of trials/contacts per trial that could be analysed.
4. Lower limb stiffness assumes the lower limb functions as a linear response spring and therefore does not consider that it is a complex system with many contributory mechanisms.

1.5 DELIMITATIONS

The following delimitations were applied:

1. Participants were restricted to high-level, able-bodied track and field athletes
2. Participants were aged between 16 and 30 years
3. Participants were free from injury and be completing full training at the time of testing
4. Assessment of lower limb stiffness was restricted to a running task at a pace reflective of the athlete's event during competition.

1.6 THESIS PRESENTATION

This thesis is presented in six chapters. Chapter Two is a narrative review of literature which presents existing knowledge surrounding lower limb stiffness, its links with performance and injury, influences of athletic training and how advancements in technology may influence how lower limb stiffness is measured. Chapter Three presents a systematic review of the longitudinal assessment of lower limb stiffness in adult athletic populations, the results of which informed the validation study. Methods applicable to the validation study are presented in Chapter Four. The validation study follows in Chapter Five and the final chapter presents a discussion of the final findings of this thesis.

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CHAPTER 2: NARRATIVE REVIEW OF LITERATURE

2.1 LOWER LIMB STIFFNESS

Lower limb stiffness describes the deformation of a body under a given force, inferring the association between the amount of leg flexion and load to which the limbs are subjected (Butler et al., 2003; Latash & Zatsiorsky, 1993). Lower limb stiffness allows the limb to increase its resistance to change under an applied load, which enhances the storage and return of elastic energy through the stretch shortening cycle (Brughelli & Cronin, 2008a). The resistance of the lower limb is dependent on the contribution of muscles, tendons, ligaments, cartilage and bone given their own individual deformation profiles, enabling lower limb stiffness to be determined at the tendon, muscle-tendon, leg and/or joint level (Brughelli & Cronin, 2008a; Butler et al., 2003).

2.1.1 Concept and Application of Lower Limb Stiffness

The concept of lower limb stiffness is reflective of Hooke's Law (Brazier et al., 2014; Brazier, Maloney, Bishop, Read, & Turner, 2017; Brughelli & Cronin, 2008a; Butler et al., 2003). Hooke's Law is defined as F=kx; where F is the force required to deform an object, k is the spring constant and x is the displacement of the spring from its equilibrium position (Butler et al., 2003; McMahon, Comfort, & Pearson, 2012). Based on this notion, the leg is often modelled as a simple spring mechanism supporting the mass of a body; referred to as the spring-mass model (Blickhan, 1989; Butler et al., 2003). Although the leg does not truly represent a physical spring this model has been effective in understanding human motion (Brazier et al., 2017).

Lower limb stiffness during human locomotion has been widely examined throughout the literature using a simple spring-mass model (Arampatzis, Schade, et al., 2001; Blickhan, 1989; Butler et al., 2003; Farley, Blickhan, Saito, & Taylor, 1991; McMahon

& Cheng, 1990). More specifically, this model has been effectively used to examine lower limb stiffness during tasks such as vertical drop jumping (Arampatzis, Schade, et al., 2001), hopping (Hobara, Kanosue, & Suzuki, 2007; Pruyn et al., 2012, 2013; Watsford et al., 2010), change of direction (Serpell, Ball, Scarvell, Buttfield, & Smith, 2014) and walking/running gait (Cavagna, Saibene, & Margaria, 1964). During tasks such as running, the interdependency of tendons, ligaments, bone and muscle all of which contribute to mechanical stiffness, can be described through the spring-mass model as it aims to quantify the relationship between the ground reaction force and deformation of the leg (Brughelli & Cronin, 2008b; Butler et al., 2003; Farley & González, 1996). The mechanical interaction of the musculoskeletal system is controlled by the stiffness of the spring during the ground contact phase of locomotion (Ferris & Farley, 1997). This model has been effectively used in the calculation of vertical and leg stiffness and has been linked to various performance outcomes (i.e. rate of force development, contact time, vertical velocity, stride length and stride frequency) (Dutto & Smith, 2002; McMahon & Cheng, 1990).

However, measurements of vertical and leg stiffness are based on the principle that the lower limb works as a global spring-mass system (Blickhan, 1989). As such, the spring-mass model does not account for the individual contribution of each joint to determine the overall stiffness of the lower limb (Latash & Zatsiorsky, 1993). Therefore, Farley, Houdijk, Van Strien, and Louie (1998) proposed the torsional spring model which analyses the stiffness of the lower limb through three torsional springs (i.e. the hip, knee and ankle). It is believed that by calculating the individual stiffness of each joint it will provide a greater understanding when it comes to describing lower limb stiffness overall, as well as provide an insight into the mechanisms which contribute to performance. For example, it has been proposed that the greatest influence on the global spring-mass system is the joint with the least amount of stiffness (Kuitunen, Ogiso, & Komi, 2011). The torsional spring model has so far been used to quantify stiffness during drop jump, hopping, walking and running tasks (Arampatzis, Schade, et al., 2001; Farley et al., 1998; Stefanyshyn & Nigg, 1998).

2.1.2 Underlying Mechanisms of Lower Limb Stiffness

Before exploring the concepts of vertical, leg and joint stiffness in more detail, it is important to consider the underlying mechanisms of lower limb stiffness. Although an advantage of assessing lower limb stiffness is its simplistic representation of a complex musculoskeletal system as a simple spring mechanism, it disregards the intricacies of what is occurring in the various neural, tendon and elastic, properties of the lower limb (Latash & Zatsiorsky, 1993).

2.1.2.1 Neural Adaptations

Optimal muscle activation and the regulation of lower limb stiffness may be influenced by neural adaptations to training (Nicol, Avela, & Komi, 2006). More specifically, the regulation of stiffness is believed to be related to structural changes of the muscle-tendon unit (MTU) and/or adjustments in mechanical behaviour (Nicol et al., 2006). In order to successfully regulate lower limb stiffness, muscle spindles relay information from the central nervous system, to accommodate for changes regarding loading (i.e. force) and muscle length (Enoka, 2008). However, under neuromuscular fatigue, the muscles ability to accommodate for these changes is decreased which places an individual at greater risk of injury given that the muscle may be subjected to loads greater than what it is able to tolerate (Nicol et al., 2006).

2.1.2.2 Tendon and Elastic Properties

Tendons are adaptable tissue that respond to stimuli and deteriorate without it (Docking & Cook, 2019). It is understood that tendons owe their stiffness to their collagen component and their viscoelasticity to their ground substance (Wang, Guo, & Li, 2012). Tendons act as a series viscoelastic component in the muscle-tendon complex and can affect the force-length/velocity relationship in the muscle (Brazier et al., 2017).

Given tendons are viscoelastic they will deform and return to their original shape in a non-linear fashion, meaning that the amount of stretch is not constant but dependent on load or time of displacement (Wang et al., 2012). The function of tendons is to store and release elastic energy and therefore tendon properties can influence rate of force development and subsequently performance (Brughelli & Cronin, 2008a). If a tendon is compliant it can result in reduced ability to generate force as it is more deformable at lower strain rates (Wang et al., 2012). It has been suggested that at low strain rate tendons are able to absorb more mechanical energy but are not as effective in transmitting mechanical loads to bone, inferring that the more compliant the tendon is, the greater the ability to store and utilise elastic energy during the stretch shortening cycle (Brughelli & Cronin, 2008a). However, it appears that there is a point of diminishing returns when the muscle-tendon complex is too compliant as it delays the force transmission from muscle to bone (Witvrouw, Mahieu, Danneels, & McNair, 2004). For example, spending too much time in the amortization phase of the stretch shortening cycle means that the energy stored during the eccentric phase dissipates as heat, and the following concentric phase will not increase muscle activity during the stretch reflex (Brazier et al., 2017).

2.1.2.3 Clinical Vs. Functional Measures of Lower Limb Stiffness

Lower limb stiffness can be divided into two categories: clinical and functional measures. The neural, tendon and elastic properties previously mentioned are often referred to as clinical measures of lower limb stiffness. More specifically, musculotendinous, tendon and passive stiffness are beneficial in providing information surrounding underlying tissue and tendon stiffness properties and are often calculated using various methods such as the free oscillation technique (Wilson, Wood, & Elliott, 1991), ultrasonography (Kubo, Kawakami, & Fukunaga, 1999) and an isokinetic dynamometer (Reid & McNair, 2004). Although valid and reliable, these measures are time consuming, have issues regarding their transportability and are potentially limited in their functional application to sport as they are isolated to a laboratory setting.

Functional measures of lower limb stiffness refer to vertical, leg and joint stiffness measures. Despite these measures not considering the intricacies of what is occurring across all of the tissues, they are more advantageous in their functional application to high performance sport. It is believed that measures of vertical, leg and joint stiffness reflect what is happening in the performance of a movement and as such will form the primary focus of this narrative review of literature.

2.1.3 Measurement and Calculation of Lower Limb Stiffness

2.1.3.1 Vertical Stiffness

Vertical stiffness is often used to calculate direct vertical movements such as hopping and vertical jumping (Butler et al., 2003). It aims to model the vertical displacement of the centre of mass (COM) and can be calculated by one of four methods (Butler et al., 2003). The first and simplest method proposed by McMahon and Cheng (1990) calculates vertical stiffness as peak vertical ground reaction force (PVGRF) divided by maximum vertical COM displacement during ground contact (Equation 1). This method has predominately been used to determine vertical stiffness using only force plate data, however COM displacement can also be determined through motion analysis systems (Brughelli & Cronin, 2008b).

Equation 1:

$$K_{vert} = \frac{F_{max}}{\Delta y}$$

Where:

F_{max} = maximum vertical force

Δy = maximum vertical COM displacement

In the second and third methods respectively, a force plate is also needed to determine vertical stiffness (Brughelli & Cronin, 2008b). These methods are based on the sinusoidal movement of the COM during ground contact. The method proposed by McMahon, Valiant, and Frederick (1987) states that vertical stiffness is equal to the mass of a body multiplied by the natural frequency of oscillation squared (Equation 2). The natural frequency of oscillation (ω) accounts for contact time and the single integration of vertical velocity.

Equation 2:

$$K_{vert} = m\omega^2$$

Similarly, the third method determines vertical stiffness as the mass of a body multiplied by the period of vertical oscillation (Cavagna, Franzetti, Heglund, & Willems, 1988) (Equation 3). The period of vertical oscillation (*P*) accounts for the total time where the vertical force is greater than body mass during ground contact (Brughelli & Cronin, 2008b). Limited research surrounding the calculation of vertical stiffness using these methods exists (Cavagna, 2006; Cavagna et al., 1988; Cavagna, Heglund, & Willems, 2005; Farley et al., 1991; McMahon et al., 1987).

Equation 3:

$$K_{vert} = m\left(\frac{2\pi}{P}\right)$$

Finally, the fourth method is the only calculation that does not require the use of a force plate. Dalleau, Belli, Viale, Lacour, and Bourdin (2004) developed a measure of vertical stiffness using contact time, flight time and body mass during hopping on a single contact mat (Equation 4). This method was found to be a valid and reliable measure of vertical stiffness when compared to a force plate, with large correlations found during maximal (r = 0.98; p <0.001) and submaximal (r = 0.94; p <0.001) hopping (Dalleau et al., 2004). As a result, this method has been used in later research (Lloyd, Oliver, Hughes, & Williams, 2009; Oliver & Smith, 2010) and may provide a feasible option for vertical stiffness to be assessed on a regular basis in the daily training environment.

Equation 4:

$$K_{vert} = \left(\frac{M \times \pi\ (T_f + T_c}{{T_c}^2\left(\frac{T_f + T_c}{\pi} + \frac{T_c}{4}\right)}\right)$$

Where:

T_f = flight time

T_c = contact time

M = body mass

While vertical stiffness does not directly measure the deformation of the lower limb, it provides researchers and sport practitioners with a fast and simple method to approximate the mechanical properties of the lower limb as a whole (Butler et al., 2003).

Vertical stiffness has most commonly been determined through the use of force plates (Cavagna et al., 1988; McMahon & Cheng, 1990; McMahon et al., 1987) and more recently contact mats (Dalleau et al., 2004), which may provide a viable field-based option for sport practitioners to assess vertical stiffness on a regular basis. However, a major limitation of vertical stiffness is that it does not account for the horizontal displacement of COM as a result of hip, knee and ankle joint flexion (Butler et al., 2003). Therefore, it may have limitations to movements such as running gait and horizontal jumping, meaning leg stiffness measures need to be considered (Butler et al., 2003; McMahon et al., 2012; McMahon & Cheng, 1990).

2.1.3.2 Leg Stiffness

Leg stiffness and vertical stiffness are often used interchangeably when the COM moves in the vertical direction (Butler et al., 2003; McMahon & Cheng, 1990). However, the measurement of leg stiffness aims to determine the compression of the leg spring and therefore a greater number of factors need to be considered (Butler et al., 2003). In Equation 5, McMahon and Cheng (1990) directly measured force as vertical ground reaction force from a force plate. Change in leg length (∆L) accounts for horizontal velocity, contact time, COM displacement and standing leg length (McMahon & Cheng, 1990). This model assumes that PVGRF occurs when the COM reaches its lowest point during mid-stance (McMahon & Cheng, 1990).

Equation 5:

$$K_{leg} = \left(\frac{F_{max}}{\Delta L}\right)$$

Change in leg length:

$$\Delta L = \Delta y + L_0(1 - Cos\theta)$$

$$\theta = sin^{-1}(\frac{ut_c}{2L_0})$$

Where:

F_{max} = maximum vertical force

ΔL = change in leg length

Δy = maximum COM displacement

L_0 = standing leg length

U = horizontal velocity

t_c = contact time

The idea by Dalleau et al. (2004) to develop a field-based stiffness measure was utilised by Morin, Dalleau, Kyröläinen, Jeannin, and Belli (2005) to develop a simple lower limb stiffness measure during running modelled around a simple sine wave. It was thought that by considering the force-time curve to be a simple sine function, vertical and leg stiffness measures could be predicted from forward velocity, leg length, body mass, flight and contact time (Brughelli & Cronin, 2008b; Morin et al., 2005) (Equation 6). This method was validated by Morin et al. (2005) when he compared stiffness scores using his method to scores from McMahon and Cheng (1990) during overground and treadmill running. The scores were reported to be acceptable (Morin et al., 2005) and as a result this method has been used in recent literature in order to provide a field-based measure of lower limb stiffness during running (Balsalobre-Fernández, Agopyan, & Morin, 2017; Ruggiero, Dewhurst, & Bampouras, 2016). The advantage of this method is that it does not require the use of a force plate to determine leg stiffness as peak vertical force is estimated using the equation below (Brughelli & Cronin, 2008b; Morin et al., 2005).

Equation 6:

1. *Estimated peak vertical force:*

$$F_{max} = mg\frac{\pi}{2}(\frac{t_f}{t_c} + 1)$$

2. *Vertical displacement of COM:*

$$\Delta y_c = \frac{F_{max} t_{c^2}}{m\pi^2} + g\frac{{t_c}^2}{8}$$

3. *Change in leg length:*

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

4. *Leg stiffness:*

$$K_{leg} = \frac{F_{max}}{\Delta L}$$

Where:

F_{max} = maximum force

m = body mass

g = gravity

t_f = flight time

t_c = contact time

L = initial leg length

v = velocity

Δy_c = maximum vertical COM displacement

Leg stiffness aims to quantify the compression of the leg spring rather than COM and as a result can be applied to horizontal movements (i.e. running). The method developed by Morin et al. (2005) enables leg stiffness to be determined without the use of a force plate and may provide sport practitioners with a feasible option to assess leg stiffness in the daily training environment. However, leg stiffness does not consider the individual contribution of each joint to the overall stiffness of the lower limb.

2.1.3.3 COM displacement

However, before discussing the calculation of joint stiffness, it is important to consider the calculation of COM displacement. COM displacement has most commonly been determined through the use of a force plate but can also be directly measured using a motion analysis system. The most common method explored throughout the literature is that of Cavagna (1975) who established that the vertical COM displacement during contact could be determined through the double integration of vertical acceleration over time. The vertical acceleration was obtained from vertical ground reaction force data divided by body mass of the participant after subtracting the gravitational acceleration (Equation 7). However, this method only accounts for changes that occur to the COM purely in a vertical direction. It does not account for COM changes during horizontal motion as it is assumed that the horizontal force is equal to zero (Butler et al., 2003), which suggests that the COM displacement may be over or underestimated (Kibele, 1999).

Equation 7:

$$a_v = \left(\frac{F_{max}}{m}\right) - g$$

Where:

a_v = vertical acceleration

F_{max} = maximum force

m = mass of a body

g = gravity

2.1.3.4 Joint Stiffness

While the most common measures of lower limb stiffness are vertical and leg stiffness, joint stiffness should also be considered, as it is needed to determine the individual contribution of each joint to the overall stiffness of the lower limb (Arampatzis, Brüggemann, & Metzler, 1999; Brazier et al., 2017; Butler et al., 2003; Farley & Morgenroth, 1999; Hobara et al., 2009). Joint stiffness has been derived using force plates and high-speed video with markers attached to appropriate body landmarks and digitised to calculate joint velocities, accelerations and displacements (Brughelli & Cronin, 2008b). The calculation of joint moments occurred through inverse dynamics (Brughelli & Cronin, 2008b). Equation 8 is the most common method used to quantify joint stiffness in the literature and is defined as the rotary version of Hooke's Law (Farley et al., 1998; Kuitunen et al., 2002; Stefanyshyn & Nigg, 1998).

Equation 8:

$$K_{joint} = \left(\frac{\Delta M}{\Delta \theta}\right)$$

Where:

ΔM = change in joint moment

$\Delta \theta$ = change in joint angle

2.1.3.5 Section Summary

Lower limb stiffness has been defined and calculated in a variety of ways in both laboratory and field based settings (Butler et al., 2003). Vertical stiffness has been used throughout the literature to assess movements such as hopping and vertical jumping, with the most commonly used protocol proposed by McMahon and Cheng (1990). On the other hand, leg stiffness has been used to quantify stiffness during tasks such as running as it accounts for the horizontal displacement of COM (Brughelli & Cronin, 2008a). Joint stiffness has been used to quantify the stiffness of a singular joint to assess the means by which individuals control stiffness to meet task demands (Brazier et al., 2017).

2.1.4 Reliability and Validity of Lower Limb Stiffness Measures

In the current literature, the reliability and validity of lower limb stiffness measures have predominantly been examined through simple vertical hopping and jump tasks. McLachlan, Murphy, Watsford, and Rees (2006) observed good inter-day reliability (ICC: 0.85-0.94, CV%: 2.7-5.0%) during normal jumps and jumps for maximum height at jump frequencies of 2.2 and 3.2Hz. Later research indicated good to moderate inter-day reliability (ICC: 0.83-0.94, CV% 19.1-21.4%) for five maximal repeat jump trials in male youths (Lloyd et al., 2009). However, these lower limb stiffness calculations were based on the use of contact and flight times from contact mats not force plate data.

Studies examining the validity and reliability of lower limb stiffness measures using force plate data have found joint stiffness during overground running (Joseph, Bradshaw, Kemp, & Clark, 2013) and hopping to be unreliable (Diggin, Anderson, & Harrison, 2016; Joseph et al., 2013). However, vertical stiffness was found to be a reliable measure of lower limb stiffness during overground running and hopping at 2.2Hz in active males (Joseph et al., 2013). Similarly, Diggin et al. (2016) found good inter and intra-day reliability of leg stiffness in thirty-two participants during single legged hopping at 1.5Hz, 2.2Hz and 3.0Hz.

2.1.5 Lower Limb Stiffness Measures and Equipment

Traditionally, measures of vertical, leg and joint stiffness have been isolated to laboratory-based settings and are most commonly determined through simple vertical hopping tasks (Farley et al., 1998; Farley & Morgenroth, 1999; Granata, Padua, & Wilson, 2002; Hobara, Inoue, Muraoka, et al., 2010; Hobara, Inoue, Omuro, Muraoka, & Kanosue, 2011; Hobara et al., 2007; Hobara et al., 2008; Hobara, Kimura, et al., 2010; Pruyn et al., 2012, 2013; Watsford et al., 2010). Hopping is the simplest locomotive task by which stiffness can be assessed and the vertical hop test has been found to be a reliable lower limb stiffness measure (Joseph et al., 2013; McLachlan et al., 2006; Pruyn, Watsford, & Murphy, 2016). Basic jump tasks (i.e. countermovement, squat, drop and horizontal jumps) have also been utilised to measure lower limb stiffness (Arampatzis, Brüggemann, et al., 2001; Arampatzis, Schade, et al., 2001; Maloney, Fletcher, & Richards, 2016; Maloney, Richards, Nixon, Harvey, & Fletcher, 2017) and vertical stiffness asymmetries (Maloney et al., 2016; Maloney et al., 2017).

Vertical hopping and jump tasks are often measured by a force plate as a direct ground reaction force measure can be obtained (Brughelli & Cronin, 2008b; Gurchiek, McGinnis, Needle, McBride, & van Werkhoven, 2017). COM displacement is usually derived from force plates but can also be determined through the use of a kinematic arm (Brughelli & Cronin, 2008b). While not commonly used throughout the literature, a kinematic arm is a device designed for the fast and simple measurement of 3D movement (Belli, Lacour, Komi, Candau, & Denis, 1995). It consists of four rigid bars linked together by three joints and allows researchers to measure COM displacement in all three planes of motion during movements such as walking and running on a treadmill (Brughelli & Cronin, 2008b). The distal end of the kinematic arm is attached to the participant while they perform the movement. The proximal end is connected to a fixed point enabling the kinematic arm to move easily through all three planes. By calculating the angle between the given bar lengths using electrical potentiometers, the immediate position of the distal end relative to the proximal end can be ascertained, allowing researchers to measure COM displacement (Brughelli & Cronin, 2008b).

Motion analysis systems also allow for displacements, joint positions and velocities to be measured (Brughelli & Cronin, 2008b). Motion analysis systems are often used in conjunction with force plates to obtain other mechanical parameters such as joint moments and joint angles (Brughelli & Cronin, 2008b; Challis, 2001). Although these measures are considered the gold standard for stiffness assessment, limited opportunity exists to measure lower limb stiffness on a regular basis in athletic populations due to restricted laboratory access, cost and difficulty with transportation (Balsalobre-Fernández, Glaister, & Lockey, 2015; Brughelli & Cronin, 2008b). Therefore, alternative options (i.e. contact mats, pressure sensors, phone applications and accelerometers) (Balsalobre-Fernández et al., 2017; Balsalobre-Fernández et al., 2015; Buchheit, Gray, & Morin, 2015; Dalleau et al., 2004; Morin et al., 2005; Morin, Jeannin, Chevallier, & Belli, 2006) need to be considered despite their limited use in published literature.

Field measures of vertical and leg stiffness have been developed using contact mats, pressure sensors and more recently Optojump technology (Dalleau et al., 2004; Morin et al., 2005; Morin et al., 2006; Ruggiero et al., 2016). As mentioned previously, the early research of Dalleau et al. (2004) established a valid and reliable measure by calculating lower limb stiffness from contact time, flight time and body mass during hopping on a single contact mat. This idea was then utilised by Morin et al. (2005) to develop a simple lower limb stiffness measure during running modelled around a simple sine wave. This method has been found to be a reliable measure of lower limb stiffness during treadmill running (Pappas, Paradisis, Tsolakis, Smirniotou, & Morin, 2014). Pappas et al. (2014) reported high intra and inter day reliability with intraclass correlations (ICC) between 0.86 and 0.99. As a result, recent research has adopted this measure using Optojump technology (Ruggiero et al., 2016). Optojump utilises infrared technology to calculate contact and flight times from interruptions in communication with the transmitted beam (Ruggiero et al., 2016). The Optojump has a high degree of validity and reliability in comparison to the reference force plate method (Ruggiero et al., 2016). However, despite its portability, Optojump is costly, limited to straight line applications, requires specific computer software and is therefore not easily accessible to many sport practitioners (Ammann, Taube, & Wyss, 2016; Brazier et al., 2017; Gindre, Lussiana, Hebert-Losier, & Morin, 2016; Ruggiero et al., 2016). Additionally, the Optojump bars are usually placed approximately 3mm above the ground, which has resulted in the overestimation of ground contact times during vertical jumping (Ammann et al., 2016). It remains unclear if contact times are under or overestimated during running tasks.

Nevertheless, recent developments in iPhone technology has presented a practical, low cost option for sport practitioners wanting to analyse lower limb stiffness in the daily training environment. Two IOS applications, more specifically “My Jump” and “Runmatic” have been designed to capture an individuals’ feet during jumping and/or running using the slow-motion capture option available on iPhone 6 and above (Brazier et al., 2017). Each application calculates contact and flight times from the video by tapping on the screen when the feet leave the ground and again when they touch the ground (Balsalobre-Fernández et al., 2017). The contact and flight times are then analysed by the application and lower limb stiffness is calculated using the method designed by Morin et al. (2005) (Balsalobre-Fernández et al., 2017). While no studies to date have directly calculated lower limb stiffness from the My Jump application, a recent study has assessed the validity and reliability of the Runmatic application when compared to Optojump for calculating leg stiffness (Balsalobre-Fernández et al., 2017). Leg stiffness was calculated from eight steps of each thirty second run of five varying speeds (i.e. 2.77, 3.33, 3.88, 4.44, 5, and 5.55 m/s) (Balsalobre-Fernández et al., 2017). Results indicated a very high level of correlation *(r* = 0.94 – 0.99) between the Runmatic application and Optojump for vertical oscillation, maximum relative force, leg stiffness, contact and flight times (Balsalobre-Fernández et al., 2017). Despite the runs being performed on a treadmill, these applications could offer a feasible alternative to gold standard stiffness measures should further evidence of validation be presented in the near future.

Finally, Global Positioning System (GPS) embedded accelerometers have recently been trialled during treadmill running to assess stride variables, running asymmetries and vertical stiffness (Buchheit et al., 2015). Small biases and large correlations were found for stride variables and vertical stiffness between the GPS embedded accelerometer and treadmill, which highlights the potential for accelerometers to assess vertical stiffness and contact time during overground running in the field (Buchheit et al., 2015). Despite promising results, a major limitation of the study was that all measures were performed on a single subject on a treadmill in the laboratory (Buchheit et al., 2015). It remains unclear whether comparable results would be reproduced in other individuals given different running patterns and anthropometric measures or whether this finding solves the issue of finding a valid and reliable stiffness measure outside of the traditional laboratory based setting (Buchheit et al., 2015). Although more studies are needed to validate this method, the potential for stiffness to be monitored in the daily training environment could provide sport practitioners with a valuable insight into the determinants of running performance and injury risk. It may also allow for the remote monitoring for coaches, medical and sport practitioners. The early detection and prevention of potential injury mechanisms may enhance athlete potential and career longevity.

2.2 EFFECTS OF LOWER LIMB STIFFNESS ON PERFORMANCE

Lower limb stiffness has known links to performance and injury risk (Butler et al., 2003). The efficient storage and return of elastic energy during stretch shortening cycle activities is dependent on a certain yet unknown amount of lower limb stiffness (Butler et al., 2003). However, it is believed that higher levels of lower limb stiffness may facilitate improved performance (Butler et al., 2003). An increase in leg stiffness has been associated with an increase in jump height (Arampatzis, Schade, et al., 2001), running velocity (Kuitunen et al., 2002), stride frequency (Farley & González, 1996) and running economy (Spurrs et al., 2003), all of which are of significance to track and field performance.

2.2.1 Jump Height and Hopping Frequency

In vertical hopping and drop jump tasks, increased lower limb stiffness has been associated with shorter contact times (Arampatzis, Schade, et al., 2001; Farley et al., 1991; Hobara et al., 2008). Hobara et al. (2008) compared power and endurance athletes and found that the power trained group displayed higher levels of leg stiffness, shorter contact and longer flight times during double leg repeat jumps across two tested jump frequencies (i.e. 1.5Hz and 3.0Hz). Shorter contact times at higher frequencies are a result of increased leg stiffness (Farley et al., 1991) which suggests that leg stiffness is directly proportional to the speed of an activity (Butler et al., 2003). This is necessary in order to resist the collapse of the lower body during the landing phase, which in turn allows for maximum energy return in the subsequent push off phase (Brughelli & Cronin, 2008a).

At the joint level, changes in vertical hopping have been primarily associated with changes in ankle and knee joint stiffness (Farley et al., 1998; Farley & Morgenroth, 1999; Hobara et al., 2007; Hobara, Kimura, et al., 2010; Hobara et al., 2009; Kim et al., 2013; Kuitunen et al., 2011). Farley and Morgenroth (1999) found increases in leg stiffness to be associated with an increase in hopping height and it was indicated that this was primarily due to the modulation of ankle joint stiffness. Later research concurred with these findings, with changes in ankle stiffness found to have the largest correlation to changes in hopping frequency (Kim et al., 2013). Contrary to these findings, it has been reported that leg stiffness adjustments for differing hopping frequencies and maximal hopping are the result of changes in knee joint stiffness (Hobara, Inoue, Muraoka, et al., 2010; Hobara et al., 2009). While knee stiffness did not correlate with vertical stiffness in the investigation conducted by Kuitunen et al. (2011), it was found to increase with greater hopping intensities and a significant relationship was found between take-off velocity and knee joint stiffness. Although these studies adopted differing methodologies, it has been suggested that ankle stiffness may be primarily associated with the modulation of contact time during hopping while knee stiffness may be closely related to the regulation of hopping height (Maloney & Fletcher, 2018).

2.2.2 Running Velocity and Stride Frequency

Leg stiffness during hopping has been positively correlated to maximal running velocity (Chelly & Denis, 2001). Athletes who display increased levels of leg stiffness during simple jump tasks have been shown to display greater levels of acceleration during running (Bret, Rahmani, Dufour, Messonnier, & Lacour, 2002). However, it has been suggested that higher stiffness levels may be detrimental for high impact reactive tasks (Walshe & Wilson, 1997). Although these studies did not directly measure the relationship between leg stiffness and running performance, studies which have produced similar findings, with an increase in lower limb stiffness associated with an increase in running velocity (Arampatzis et al., 1999; Kuitunen et al., 2002) and stride frequency (Farley & González, 1996).

Increased lower limb stiffness is of particular interest in maximal effort sports such as sprinting as it has been associated with a decrease in contact time, stride length and vertical COM displacement (Brughelli & Cronin, 2008a; Farley & González, 1996; Kuitunen et al., 2002). This is beneficial for sprinting performance as a decrease in leg stiffness has been associated with an increase in stride length (Farley & González, 1996). This decrease can negatively influence running velocity if stride length sacrifices stride frequency, further supporting the notion that the level of stiffness required is dependent on task demands (Kuitunen et al., 2002).

The individual joint contributions of the lower limb have also been examined and associated with an increase in running velocity (Butler et al., 2003). Studies exhibiting forefoot striking patterns have found increases in running velocity to be associated with an increase in knee stiffness (Arampatzis, Schade, et al., 2001; Kuitunen et al., 2002). However, studies examining rearfoot landing patterns have reported that an increase in ankle stiffness is directly related to an increase in running velocity (Arampatzis et al., 1999; Laughton, McClay Davis, & Hamill, 2003) thereby suggesting that further investigation of individual joint contributions to overall stiffness and performance benefits are needed.

2.2.3 Running Economy

In addition to maximal effort sports such as sprinting, stiffness has been linked to running economy, which is of particular relevance to distance running performance (Butler et al., 2003). A study examining steady state oxygen consumption and stiffness found an increase in musculotendinous stiffness to be associated with running economy (Spurrs et al., 2003). An increase in vertical stiffness has also been associated with improved running economy during normal running gait when compared to ‘Groucho’ running (a greater degree of knee flexion) (McMahon et al., 1987). Similarly, it has been suggested that fatigue may contribute to decreased vertical and leg stiffness resulting in decreased running economy (Dutto & Smith, 2002). Increased lower limb stiffness is believed to be beneficial to running economy as it allows more effective storage of elastic energy during the eccentric phase of the stretch shortening cycle thereby resulting in lower energy expenditure (Kuitunen et al., 2002).

2.2.4 Section Summary

Overall, stiffness measures are particularly relevant for track and field performance as they are highly related to factors such as increased jump height (Arampatzis, Schade, et al., 2001), running velocity (Kuitunen et al., 2002), stride frequency (Farley & González, 1996) and running economy (Spurrs et al., 2003). Simple jump measures have been found to be significantly different between power and endurance athletes (Hobara et al., 2008) with the level of stiffness required to perform a task dependent on task demands (Kuitunen et al., 2002). However, few studies have focused on lower limb stiffness in track and field athletes during tasks reflective of their training and competition background.

2.3 EFFECTS OF LOWER LIMB STIFFNESS ON INJURY RISK

While many accept that increased lower limb stiffness improves performance (Arampatzis, Brüggemann, et al., 2001; Avela & Komi, 1998; Farley & González, 1996; Kuitunen et al., 2002; Spurrs et al., 2003), others argue that too much or too little stiffness may increase an individual's risk of injury (Butler et al., 2003; Granata et al., 2002; Williams et al., 2004; Williams et al., 2001). Although further research is needed to explore and understand the direct relationship between lower limb stiffness and injury, higher levels of stiffness have been associated with bone related injuries while lower levels of stiffness have been linked with soft tissue injuries (Butler et al., 2003).

2.3.1 Bone Related Injuries

An increase in lower limb stiffness has been associated with reduced excursions of the lower limb and an increase in peak ground reaction force (Butler et al., 2003). This combination has been suggested to increase loading rates thereby resulting in an increased risk of bone related injuries (Butler et al., 2003). Although strong evidence supporting this association is lacking, prospective studies have investigated this finding, which provide some support. For example, previous authors studied the structure of an individual's foot on lower limb stiffness mechanics and injury patterns (Powell, Paquette, & Williams, 2017; Williams et al., 2004). It was found that high arch runners exhibited greater levels of lower limb stiffness than low arch runners (Powell et al., 2017; Williams et al., 2004). This increase in lower limb stiffness may be due to an increase in peak ground reaction force as the high arch runners are unable to attenuate shock (Williams et al., 2001). Given that high arch runners have been associated with an increased incidence of bone related injuries such as tibial stress fractures and lateral ankle sprains (Powell et al., 2017; Williams et al., 2001), it is plausible to consider that high arch runners may be at higher risk of injury due to increased lower limb stiffness.

2.3.2 Soft Tissue Injuries

Conversely, too little stiffness has been associated with soft tissue injuries and is believed to be the result of excessive joint motion (Butler et al., 2003). Low arch runners were found to experience more soft tissue injuries than their high arch counterparts because of decreased leg stiffness (Williams et al., 2004). Granata et al. (2002) found lower limb stiffness to vary between genders, with females displaying an impaired ability to regulate stiffness. However, later research found that when body weight was normalised, gender differences in vertical stiffness were eliminated during the same task (Padua, Carcia, Arnold, & Granata, 2005). Despite this, a potential contributing factor for such differences was joint stability, with females displaying greater levels of joint laxity than their male counterparts (Granata et al., 2002). Increased joint laxity has been associated with decreased stiffness levels (Farley et al., 1998), which in turn offers a possible explanation as to why females are at greater risk of injury. Under fatigue, females have also been found to rely more on the ankle than knee musculature to control vertical stiffness than men (Padua et al., 2006). It has been theorised that an ankle dominant strategy may reduce knee joint stability, which may provide an explanation as to why females have an elevated risk of anterior cruciate ligament injuries (Granata et al., 2002; Padua et al., 2006; Padua et al., 2005).

2.3.3 Longitudinal Injury Links

Despite links between lower limb stiffness, performance and injury risk; longitudinal assessment of lower limb stiffness and its impact on performance and injury remains scarce. As a result, the following associations are based on cross-sectional/retrospective study designs.

Pruyn et al. (2012) studied the bilateral differences in leg stiffness and its contribution to non-contact lower limb soft tissue injuries in AFL players. It was concluded that players with a moderately high bilateral difference in leg stiffness may be more prone to non-contact, soft tissue injuries in the lower limb (Pruyn et al., 2012). These findings were in agreeance with the earlier research of Watsford et al. (2010) who found that higher bilateral leg stiffness may be a factor in sustaining a non-contact, soft tissue hamstring injury in AFL footballers. However, later research investigating the same sporting population disputed such findings with no relationship found between vertical stiffness and lower limb muscle strains (Serpell, Scarvell, et al., 2014). Although each of these studies employed simple jump tasks for stiffness assessment, Serpell, Scarvell, et al. (2014) only analysed a single jump which places questions around the accuracy of the data obtained. A study assessing data reduction on inter-trial variability of lower limb stiffness suggested that one contact is insufficient with at least three consecutive contacts considered optimal (Moresi, Bradshaw, Greene, & Naughton, 2015).

The regularity by which stiffness is currently assessed in athletic populations has predominately been limited to pre and/or post season measures (Nagahara & Zushi, 2017; Watsford et al., 2010). The relevance of existing knowledge and its practical application for coaches and athletes is limited due to the irregularity of stiffness measures. Therefore, on-going field-based measures of lower limb stiffness are needed to potentially identify stiffness changes that may not have been previously seen, which may impact on performance and injury.

A recent study by Pickering Rodriguez, Watsford, Bower, and Murphy (2017) examined the relationship between lower limb stiffness and non-contact injuries in sub-elite and elite netballers throughout a domestic competition season. The results revealed no significant difference in vertical stiffness between the injured and non-injured groups which also contrasts the earlier findings of Watsford et al. (2010). Although lower limb stiffness has been found to vary between genders (Granata et al., 2002), it is unknown to what extent lower limb stiffness is affected by different training modalities, match play and playing position. Therefore, further research is needed to examine the longitudinal effects of lower limb stiffness in varying athletic populations using tasks reflective of their training and competition background to further understand the link between lower limb stiffness, performance and injury.

2.3.4 Section Summary

Research suggests that an optimal level of stiffness exists for maximisation of performance and injury prevention (Arampatzis et al., 1999; Kuitunen et al., 2002; McMahon & Cheng, 1990). However, the quantification of this optimal level of stiffness presents a challenge as lower limb stiffness can place individuals at higher risk of injury

(Pruyn et al., 2012; Watsford et al., 2010). Higher levels of stiffness can result in an increased risk of bone related injuries, due to the associated rapid transmission of force from muscle to bone, while lower levels of stiffness may increase an individual's risk of soft tissue injuries (Butler et al., 2003). With increased stiffness appearing beneficial to performance, but also placing an athlete at increased risk of bone related injuries, an optimal level of stiffness for maximisation of performance and injury prevention may exist (Butler et al., 2003). However, these theoretical stiffness limits remain unknown in athletic populations.

2.4 INFLUENCES OF ATHLETIC TRAINING ON LOWER LIMB STIFFNESS

Despite theoretical stiffness limits being unknown, it is thought that athletic training is one factor that may potentially influence an individual's level of stiffness. Although limited research looking at the influence of training on vertical, leg and joint stiffness during sport specific tasks exists (Millett, Moresi, Watsford, Taylor, & Greene, 2017), several studies have detailed changes in lower limb stiffness following plyometric (Burgess, Connick, Graham-Smith, & Pearson, 2007; Kubo et al., 2007; Spurrs et al., 2003) and resistance training (Burgess et al., 2007; Kubo et al., 2007).

2.4.1 Plyometric Training

It is widely known that plyometric training allows athletes to gain performance benefits such as improved running economy (Spurrs et al., 2003) and increased jump height (Burgess et al., 2007; Kubo et al., 2007; Toumi, Best, Martin, & Poumarat, 2004). These benefits are believed to be the result of neuromuscular adaptations following plyometric training. Kyröläinen, Komi, and Kim (1991) reported that four weeks of plyometric training increased the pre-activity of lower limb muscles, which in turn led to improved intramuscular coordination and increased musculotendinous stiffness. It has been suggested that higher levels of stiffness in the muscles of the lower limb provide an advantage for participants undertaking stretch shortening cycle exercises as greater amounts of elastic energy can be stored and reused (Komi, 2003). This was supported by the later research of Kubo et al. (2007) who found that stored elastic energy and Achilles tendon elongation significantly increased following plyometric training, which in turn lead to improved stretch shortening cycle jump performance.

2.4.2 Jump Training Programs

Jump training programs incorporating plyometric exercises have not only been associated with performance benefits but also reducing an individual's risk of injury (Hewett, Lindenfeld, Riccobene, & Noyes, 1999; Hewett, Stroupe, Nance, & Noyes, 1996). Hewett et al. (1996) introduced a jump training program incorporating plyometric exercises that aimed to teach individuals to land 'softer', ultimately allowing them to achieve optimal knee flexion during landing. After the six-week intervention, a reduction in PVGRF was evident (Hewett et al., 1996). Further exploration of this concept by Hewett et al. (1999) found that female athletes who participated in the jump training program had a significantly lower risk of injury occurrence because of their ability to optimise knee flexion, which in turn resulted in reduced PVGRF. The reduction in PVGRF was associated with a decrease in lower limb stiffness suggesting that an individual may be able to intentionally control lower limb stiffness and possibly reduce their risk of injury (Hewett et al., 1999). These findings were supported by the later research of Chimera, Swanik, Swanik, and Straub (2004) who found that plyometric training may reduce the incidence of injury by lowering joint stiffness in the lower limb. Therefore, it can be concluded that appropriate jump landing interventions may elicit positive changes in biomechanical technique by subsequently reducing joint loading.

2.4.3 Verbal Instructions

Verbal cues given to participants such as "jump as high as you can" or "jump high a little faster than your previous jump" have also been found to influence lower limb stiffness (Arampatzis, Brüggemann, et al., 2001, p. 953; Arampatzis, Schade, et al., 2001, p. 356). Both studies reported that leg stiffness increased with shorter contact times, which puts forward the idea that lower limb stiffness can be intentionally controlled through variations in contact time (Arampatzis, Brüggemann, et al., 2001; Arampatzis, Schade, et al., 2001). Effective coaching cues are believed to result in higher muscle activation which in turn results in greater ground reaction force, shorter contact times and reduced COM displacement, all of which influence leg stiffness (Arampatzis, Brüggemann, et al., 2001).

2.4.4 Heavy Resistance Training

Along with plyometric training and verbal instructions, heavy resistance training has also been shown to affect lower limb stiffness. It is a common training method used by athletes not only to increase strength but also enhance power and movement speed whereby "relatively heavy loads (80-90% maximum) are lifted for relatively few repetitions (4-8 repetitions)" (Wilson, Newton, Murphy, & Humphries, 1993, p. 1279). Resistance training using heavy load has been associated with increased stiffness of the MTU (Kubo et al., 2007; Kubo, Yata, Kanehisa, & Fukunaga, 2006). Tendon stiffness was measured using ultrasonography (Kubo et al., 2007; Kubo et al., 2006). A stiffer MTU is thought to increase rate of force development (Brughelli & Cronin, 2008b). This is believed to be beneficial for maximum effort sports such as sprinting as the ultimate goal is to cover a certain distance in the least possible time (Bezodis, Kerwin, & Salo, 2008) inferring that the rapid production of forces in the lower limb is crucial to high performance. However, no research to date has examined the influence of resistance training on vertical, leg and joint stiffness during human running (Brughelli & Cronin, 2008b). Further research is needed to examine how strength-induced changes in lower limb stiffness affect sprint performance.

2.4.5 Section Summary

It is evident that the majority of research examining the influence of athletic training on lower limb stiffness has utilised simple jump tasks (Burgess et al., 2007; Kubo et al., 2007; Kubo et al., 2006; Kyröläinen et al., 1991). Although the extent to which training influences lower limb stiffness during running is yet to be determined, it is thought that a combination of plyometric and resistance training may contribute significantly to improvements in explosive leg power and dynamic athletic performance (Markovic, Jukic, Milanovic, & Metikos, 2007). Further research is needed to investigate the interaction of training load and changes to the way an athlete deals with loading specific to their sport to potentially enable early identification of injury risk and performance maximisation. With advancements in technology, this is now thought to be a possibility in the daily training environment.

2.5 LABORATORY VS FIELD: ADVANCEMENTS IN TECHNOLOGY

Traditionally, the analysis of human movement and measures of lower limb stiffness have been isolated to laboratory-based settings (Lee, Mellifont, & Burkett, 2010; Little et al., 2013). The artificial nature of a laboratory setting is thought to influence natural running gait patterns as well as limit the scope of data collected (Hood et al., 2012; Lee, Mellifont, et al., 2010; Little et al., 2013; Mayagoitia, Nene, & Veltink, 2002). With advancements in technology, data collection equipment is becoming smaller, portable and more affordable allowing the assessment of human movement to occur in a variety of settings (Fong & Chan, 2010; Higginson, 2009; Winter, Lee, Leadbetter, & Gordon, 2016). Human movement analysis can show more transferable application to executions of day to day environments with the advancements in these technologies thus, uncovering more longitudinal information (Lee, Sutter, Askew, & Burkett, 2010; Norris, Anderson, & Kenny, 2013).

2.5.1 Inertial Measurement Units

Inertial measurement units allow acceleration, heading and orientation to be captured and data logged in a small unit (Hood et al., 2012; Seel et al., 2014). The ability to identify certain gait parameters from inertial measurement units has been known for some time (McCamley, Donati, Grimpampi, & Mazzà, 2012). However, validation of inertial measurement units has so far been limited to stride parameters, contact times and vertical displacement (Lee, Sutter, et al., 2010). As a result, there has been limited use of inertial measurement units in obtaining a lower limb stiffness measure, with only one study reporting the use of an accelerometer. Hobara, Inoue, Gomi, et al. (2010) used a biaxial accelerometer attached to the right heel of each athlete to determine flight and ground contact times of each step during a 400m sprint. All the other required variables for the stiffness model were obtained through anthropometric measures and video analysis (Hobara, Inoue, Gomi, et al., 2010) which highlights the limited impact accelerometers and inertial measurement units have so far had in obtaining a lower limb stiffness measure.

While validation of inertial measurement units have so far been limited to stride parameters, contact times and vertical displacement, a recent conference paper explored the validation of inertial measurement units against a Laveg laser to determine velocity measures at each 10m split of a 100m sprint (Parrington et al., 2016). Although further validation is required given the small sample size, results showed promise with the inertial measurement unit data and Laveg laser strongly correlated (Parrington et al., 2016). However, it should be noted that only a medium correlation was evident for the first 10m split (Parrington et al., 2016). Should this be appropriately validated in the future, inertial measurement units may provide a valuable tool for assessing athletic performance as well as provide a feasible option for sport practitioners, coaches and athletes when it comes to collecting various performance measures.

2.5.2 Motion Analysis Systems Vs. Inertial Measurement Units

3D motion analysis is considered the laboratory gold standard for the study of human movement in biomechanical settings (Cuesta-Vargas, Galán-Mercant, & Williams, 2010; Hood et al., 2012; Iosa, Picerno, Paolucci, & Morone, 2016). These systems utilise reflective markers for the measurement of human movement thus allowing the tracking of motions in three dimensions (Cuesta-Vargas et al., 2010; Iosa et al., 2016). However, motion analysis systems have limited capture volume, require specialised laboratories, involve long post-processing times and are therefore not convenient for large populations or studies that involve long lasting movements (Brughelli & Cronin, 2008b; Iosa et al., 2016).

Recent research comparing the use of inertial measurement units and motion analysis systems have produced favourable and consistent results. Mayagoitia et al. (2002) found that body mounted sensors incorporating accelerometers and gyroscopes attached to the frontal medial aspect of the shank and thigh were accurate at five different walking speeds when compared with a motion analysis system. Later research by Lee, Mellifont, et al. (2010) reported an agreement between an inertial sensor placed on the sacrum and an infrared camera system for step, stride and stance measures during running. Large correlations, small bias and standard error were found across all of the running gait variables, demonstrating a strong relationship between both methods (Lee, Mellifont, et al., 2010). Lee, Sutter, et al. (2010) also found near perfect correlations for COM vertical accelerations between both methods during treadmill running. These findings highlight the agreement between inertial measurement units and motion analysis systems, potentially enabling the assessment of human movement outside traditional laboratory settings. However, the reliability and accuracy of inertial measurement units has been found to be dependent on the task being performed and site to which they are attached (Cuesta-Vargas et al., 2010). Therefore, further research and development of inertial measurement units are needed to improve their use in the analysis of human movement.

2.5.3 Force Plate Vs. Inertial Measurement Units

Along with motion analysis systems, force plates are also considered the laboratory gold standard when it comes to understanding human movement (Challis, 2001; Hood et al., 2012). Despite their vast use throughout the literature, a major limitation of using force plates to measure athletic performance are issues surrounding their transportability (Brughelli & Cronin, 2008b). Often force plates are expensive, fixed in position and thereby limit the amount of data that can be obtained (Brughelli & Cronin, 2008b; Hood et al., 2012). Not only do they limit the amount of data obtained, they may also influence an athlete's natural movement pattern as athletes may alter their natural gait in order to strike the force plate (Challis, 2001). Therefore, recent research has looked at the comparison of inertial measurement units and ground mounted force plates.

Patterson et al. (2016) looked at the validation of three inertial measurement unit locations in comparison to a force plate during normal paced walking in healthy adults. The step times from all three inertial measurement placements (i.e. trunk, shank and feet) were all found to be valid measures (Patterson et al., 2016). However, the inertial measurement units placed on the feet were found to be the most accurate in comparison to the force plate (r = 0.991) due to less attenuation of ground reaction forces (Patterson et al., 2016). Furthermore, later research by Gurchiek et al. (2017) investigated the use of a sacrum-worn inertial measurement unit to estimate 3D ground reaction forces during sprint starts and change of direction tasks. The results revealed a good level of agreement between vertical force and a poor level of agreement between the lateral force components (Gurchiek et al., 2017). Similarly, Raper et al. (2018) designed a protocol to measure ground reaction force during running using a single inertial measurement unit mounted on the medial tibia. The results revealed that the inertial measurement unit underestimated the ground reaction force when compared to the force plate. This was believed to be the result of a delay between the peak of exerted force and peak in acceleration (Raper et al., 2018). While the results of these studies may have increased the scope for the use of inertial measurement units outside the traditional laboratory setting, further research is needed to reliably fill the gap between laboratory and field-based measures.

2.5.4 Section Summary

Although the analysis of human movement has traditionally been isolated to laboratory based settings, it is not without limitations (Higginson, 2009). Recent advancements in technology has allowed accelerometers, gyroscopes and magnetometers to be incorporated and data logged in small, lightweight, portable units (Higginson, 2009). These advancements are believed to allow data collection to occur in unconstrained environments, meaning tracking and monitoring of athletes in the daily training environment is now a possibility. While the limitations of gold standard equipment (i.e. force plates, motion analysis systems) are likely driving factors in the development of new technologies such as inertial measurement units, it is important that all new devices are found to be valid and reliable otherwise data may become detrimental to an athlete's performance. It is imperative that measures be practical and accurate before adopted for decision making on training and performance outcomes/modifications as well as for the monitoring of stiffness in the daily training environment to effectively assess stiffness and injury longitudinally.

2.6 CONCLUSION

The review of literature highlights the historical perspectives, calculations of lower limb stiffness and its link with performance and injury. It is apparent that much of the research surrounding lower limb stiffness has primarily focused on simple jump tasks in laboratory based-settings with differences reported between power and endurance athletes (Hobara et al., 2008). Currently, longitudinal measures of lower limb stiffness and its effects on performance and injury are performed on an irregular basis and as such have only provided sport practitioners with a brief insight into such links. In addition, research investigating the interaction of training load and the way individuals respond to loading (i.e. leg stiffness) is in its infancy. However, due to recent advancements in technology, measurement of leg stiffness may be a possibility with inertial measurement units. The application of this novel technology is believed to make data collection feasible in the daily training environment as well as provide a proactive approach to the management of athlete performance and injury (i.e. reduce lost training time). It is thought that if this technology can be appropriately validated, it may potentially allow for more regular, long-term stiffness tracking and the interaction of training load in a functional, applied manner.

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CHAPTER 3: SYSTEMATIC REVIEW

The initial focus of this research was to implement a novel field-based measure to quantify lower limb stiffness in high level track and field athletes over a full preparation period (i.e. 6 – 9months). Therefore, a systematic review surrounding the longitudinal assessment of lower limb stiffness in adult athletic populations was conducted at the conception phase of this research. While the original focus of this research was longitudinal tracking, it became apparent via this review that there was a need for field-based measures of lower limb stiffness and therefore the focus of this research shifted. It is important to note that this review still addresses common themes explored throughout the narrative review of literature. More specifically, it identifies that measurements of lower limb stiffness have been isolated to laboratory-based settings and are most commonly determined through vertical hopping and jump tasks. It also highlights the limited available knowledge surrounding the influence of athletic training on lower limb stiffness. It is hoped that the gaps highlighted throughout this review, contribute to the future direction of this research in terms of continuing to develop a field-based measure to determine longitudinal stiffness changes during tasks specific to an athlete's sport/chosen event.

3.1 ABSTRACT

The relationship between lower limb stiffness, athletic performance and injury is of significant interest to the wider sporting and research community. However, much of the research to date has examined this relationship through cross sectional research with long-term stiffness changes in athletic populations remaining relatively unknown. The databases of CINAHL complete, SPORTDiscus, MEDLINE complete, Web of Science and Cochrane Library were searched from their earliest record to 4th March 2019. Included studies were longitudinal/observational studies in which the data collection period was greater than three months and where an applied measure of lower limb stiffness was obtained (i.e. vertical, leg and/or joint stiffness). Only six longitudinal studies were found to examine lower limb stiffness and its links with performance and injury in athletic populations. However, these studies predominately utilised simple vertical hopping and jumping tasks for stiffness assessment, provided infrequent stiffness measures and mainly focused on the specific sporting population of AFL players. Further investigations are needed to examine the longitudinal assessment of lower limb stiffness in varying athletic populations using more sport specific measures at regular intervals to better understand the prospective links between lower limb stiffness, performance and injury.

Keywords: Injury, athletic performance, lower extremity, long-term

3.2 INTRODUCTION

Measures of lower limb stiffness model the leg as a simple spring mechanism, which attenuates the forces applied to the body (Butler et al., 2003). The compliance or resistance of this 'leg spring' can be quantified using force and displacement measures providing a lower limb stiffness measure (Butler et al., 2003). The integration of muscles, ligaments, tendons, cartilage and bone contribute to the overall musculoskeletal system acting like a spring (Brughelli & Cronin, 2008a).

Lower limb stiffness has been associated with performance and injury risk (Butler et al., 2003). Several studies have examined this relationship through the use of vertical hopping and jumping tasks (Arampatzis, Schade, et al., 2001; Hobara et al., 2008; Pruyn et al., 2012; Watsford et al., 2010) with studies focusing on running gait measures less common throughout the literature (Farley & González, 1996; Kuitunen et al., 2002). However, the few studies focusing on running measures have produced comparable results, with such findings being of significance to sports performance. An increase in leg stiffness has been associated with an increase in jump height (Arampatzis, Schade, et al., 2001), running velocity (Kuitunen et al., 2002), stride frequency (Farley & González, 1996) and running economy (Spurrs et al., 2003). However, the literature suggests there is an optimal theoretical range of stiffness that will improve performance without increasing the risk of injury. Studies have found that greater levels of stiffness are associated with bone related injuries due to the associated rapid transmission of force from muscle to bone (Butler et al., 2003; Hobara et al., 2008), while lower levels of stiffness have been linked to soft tissue injuries (Butler et al., 2003). However, these associations appear primarily based on cross-sectional research, with limited longitudinal studies examining such links.

To improve athletic performance and reduce injury risk, optimal levels of functional stiffness such as vertical, joint and leg stiffness are essential (Brughelli & Cronin, 2008a). Optimisation of leg stiffness can allow the limb to increase its resistance to change under an applied load, enhancing the storage and return of elastic energy necessary for peak performance. However, these measures have been predominately isolated to laboratory-based settings. Force plates have been widely used throughout the literature because a direct ground reaction force measure can be obtained (Brughelli & Cronin, 2008b). COM displacement is usually derived using force plate measures but can also be determined from video or a motion analysis system (Brughelli & Cronin, 2008b; Butler et al., 2003). Force plates are often used in conjunction with a fixed motion analysis system to obtain other mechanical parameters such as joint moments and joint angles (Brughelli & Cronin, 2008b) which are necessary for the calculation of joint stiffness.

As most measures of lower limb stiffness have been measured using force plates and/or motion analysis, regular stiffness assessment in athletic populations presents a challenge due to restricted laboratory access, cost and difficulty with transportation (Brughelli & Cronin, 2008b). Therefore, longitudinal assessment of lower limb stiffness and its impact on performance and injury remains scarce. Two longitudinal studies investigating AFL footballers found leg stiffness and limb asymmetry to be linked with potential injury risk (Pruyn et al., 2012; Watsford et al., 2010). However, later research focusing on the same sporting population contrasted such findings with no relationship found between vertical stiffness and lower limb muscle strains (Serpell, Scarvell, et al., 2014). While these studies established conflicting results, it should be noted that they adopted differing methodologies with two of the studies assessing lower limb stiffness through repeat hopping (Pruyn et al., 2012; Watsford et al., 2010) and one study assessing stiffness through a single jump (Serpell, Scarvell, et al., 2014). Despite links between acute stiffness changes and performance, longitudinal changes to stiffness and subsequent links with performance and injury remain unclear in athletic populations.

Additionally, limited research has been undertaken to understand the longitudinal effects of training on lower limb stiffness. Following a ten week training program, Hunter and Marshall (2002) found that the power trained group displayed increased leg stiffness during countermovement jumps but decreased leg stiffness during all of the drop jump conditions. Later research investigating the effects of plyometric and weight training on jump performance, found an increase in ankle stiffness during countermovement and drop jumps to be associated with twelve weeks of plyometric training (Kubo et al., 2007). Similarly, Nagahara and Zushi (2017) found an increase in ankle and vertical stiffness during maximal sprinting following a six-month training period. Although these findings are based on different types of training and tasks, the majority of these studies only tested stiffness prior to and following the completion of the training period, which highlights the limited research surrounding the regular effects of training on lower limb stiffness.

The regular monitoring of lower limb stiffness in athletic populations is thought to be important for optimising performance and managing potential injury risk (Butler et al., 2003). Therefore, the purpose of this study was to systematically review and compare existing knowledge surrounding populations, methodologies and findings from investigations already conducted and provide future direction into the longitudinal monitoring of lower limb stiffness in athletic populations.

3.3 METHODS

3.3.1 Search Strategy

Electronic database searches were performed in CINAHL complete, SPORTDiscus, MEDLINE complete (Via EBSCO), Web of Science and the Cochrane Library from their earliest record to 4th March 2019. The database searches were performed using the following three groups of keyword items: (1) leg stiffness, vertical stiffness, spring mass, leg spring, (2) sports performance, athletic performance, leg injury, lower limb injury, (3) professional, elite, athlete, high-level. Terms in each group of keywords were combined using the Boolean operator “OR” and then the three groups were combined with an “AND”.

3.3.2 Study Inclusion and Exclusion

Studies were included if they met all the following criteria: (1) the literature was written in English, (2) participants were human (healthy/adult/physically-mature populations), (3) the study focused on the lower extremity, (4) an applied measure of stiffness was obtained (i.e. vertical, leg and/or joint stiffness) and (5) the study was a longitudinal/observational study in which the data collection period was greater than three months. Three months was deemed by the researchers to be the minimum amount of time needed to qualify a study as longitudinal. No limit was placed on the duration or frequency of stiffness measures used.

Experimental, intervention and descriptive studies that assessed the influence of bracing, shoes, taping, prosthesis, orthoses, genetics, rehabilitation and the physiological effects (i.e. female menstrual cycle) on stiffness were excluded. Unpublished studies, review papers, conference abstracts and studies that primarily focused on clinical measures of stiffness (i.e. muscle, tendon, passive stiffness) were also excluded.

3.3.3 Data Extraction

Two reviewers (TW, MM) developed the review protocol and determined the inclusion and exclusion criteria. Studies that could not be eliminated by title and abstract were retrieved and independently assessed by TW. Reference lists of studies retrieved for inclusion were also manually searched by TW to identify any additional eligible papers. Data relating to study design, participant characteristics, methodology and frequency of stiffness assessment were extracted for analysis.

3.4 RESULTS

3.4.1 Identification and Selection of Studies

The initial search strategy produced 630 records. Following the removal of duplicates, 452 relevant studies remained of which 430 were excluded based on the inclusion/exclusion criteria during the initial screening of titles and abstracts. Twenty-two articles were included for full text review with 13 excluded due to their cross-sectional study design and 3 excluded due to their repeated measures experimental approach being less than three months. A total of 6 studies were identified for review (Figure 3.1).

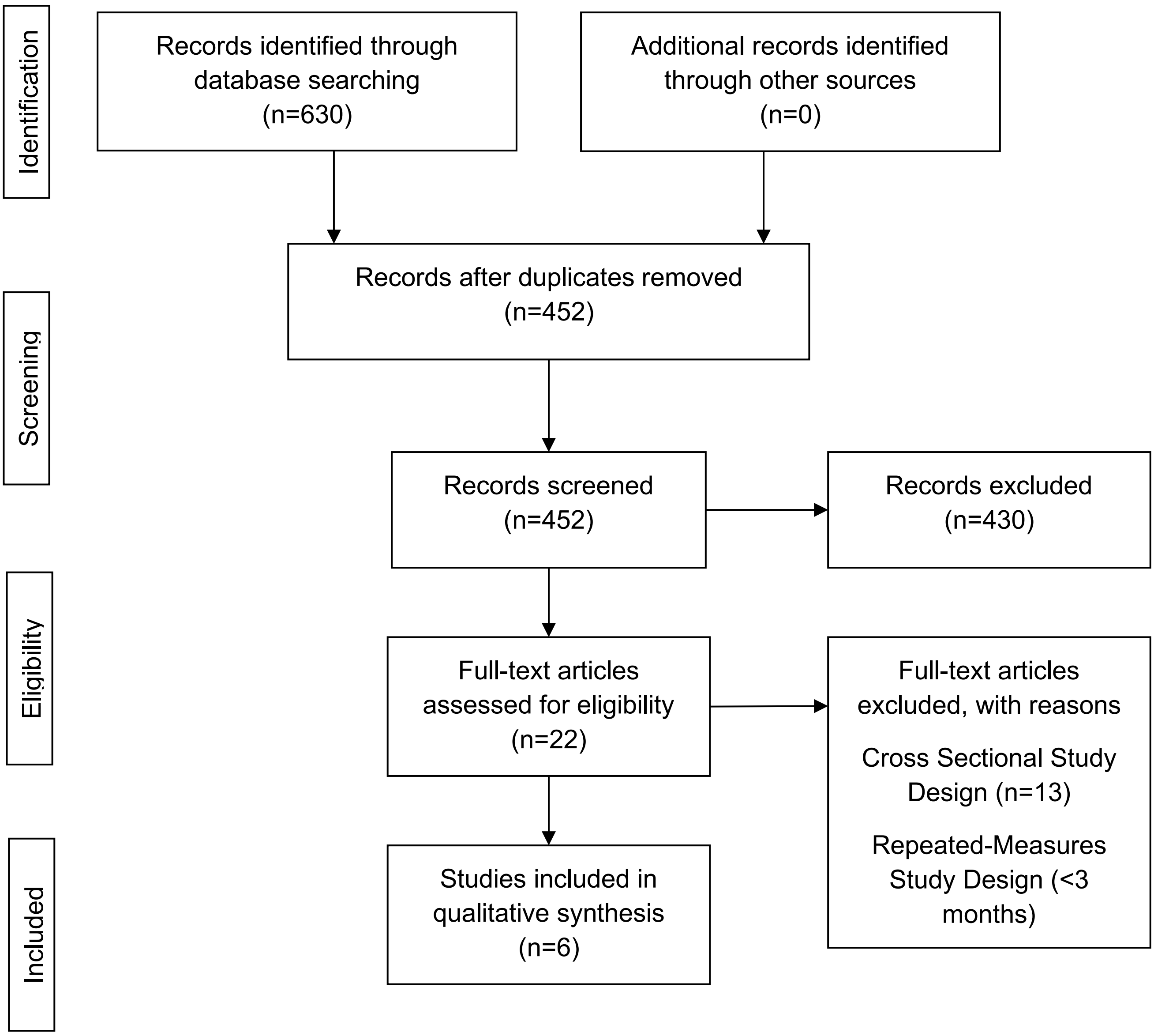

Figure 3.1 PRISMA flowchart for identification and inclusion of relevant studies

3.4.2 Study Characteristics

It is evident from the inclusion criteria that all studies employed a longitudinal study design. There were three prospective cohort studies (Pickering Rodriguez et al., 2017; Pruyn et al., 2012, 2013), one retrospective cohort study (Serpell, Scarvell, et al., 2014), one case control study (Watsford et al., 2010) and one within subject repeated measures study (Nagahara & Zushi, 2017) included for review. Four of the included studies investigated the relationship between leg/vertical stiffness and injury as their primary aim (Pickering Rodriguez et al., 2017; Pruyn et al., 2012; Serpell, Scarvell, et al., 2014; Watsford et al., 2010). One study aimed to determine how vertical, leg and joint stiffness were altered in maximal speed sprinting performance after a six month training program (Nagahara & Zushi, 2017). The other remaining study reported an intention to investigate the interaction between leg stiffness and training load (Pruyn et al., 2013). Three of the six studies reported a hypothesis suggesting that a relationship between the respected investigated variables would exist (Pruyn et al., 2012, 2013; Watsford et al., 2010), while one study hypothesised that different stiffness profiles would exist between injured and non-injured players (Pickering Rodriguez et al., 2017). However, only two studies reported a directional hypothesis suggesting that higher levels of leg stiffness would be related or a risk factor for a non-contact soft tissue injury (Pruyn et al., 2012; Watsford et al., 2010).

3.4.3 Participant Characteristics

Extracted participant characteristics are shown in Table 3.1. One study recruited nine well-trained male track and field athletes (one sprinter, one decathlete, two jumpers and five pole vaulters) from a university athletic club (Nagahara & Zushi, 2017). The population recruited in four of the six studies were professional Australian Rules football players (Pruyn et al., 2012, 2013; Serpell, Scarvell, et al., 2014; Watsford et al., 2010) The sample size ranged from 25 to 166 participants. The mean age, height and weight of participants ranged from 20.5 to 27 years, 83.1 to 88.4 kg and 185 to 189.8 cm respectively. All four studies failed to report gender. However, it is assumed that all participants were male as there were no female professional Australian Rules football players at the time of publication.

The remaining study recruited 10 elite and 19 sub-elite female netball players (Pickering Rodriguez et al., 2017). Of the 29 players, 11 were shooters, 10 were defenders and 8 were centre court players. The mean age, height and weight for the whole group was 24.1 years, 72.4 kg and 178.0 cm respectively.

Table 3.1 *Participant Characteristics*

Authors	Population	Sample	Age (yrs)	Body Mass (kg)	Height (cm)
Nagahara et al. 2017	Track and Field Athletes	n = 9	19.4 ± 1.0	65.9 ± 3.9	174.0 ± 0.05
Pickering Rodriguez et al. 2017	Elite and Sub-Elite Netballers	<u>Elite</u> NON-INJ: n = 5 INJ: n = 5 <u>Sub-elite</u> NON-INJ: n = 14 INJ: n = 5	<u>Elite</u> NON-INJ: 25.9 ± 3.7 INJ: 25.8 ± 4.4 <u>Sub-elite</u> NON-INJ: 23.4 ± 2.7 INJ: 22.5 ± 2.0	<u>Elite</u> NON-INJ: 75.2 ± 5.4 INJ: 75.4 ± 6.8 <u>Sub-elite</u> NON-INJ: 71.0 ± 6.0 INJ: 70.6 ± 5.9	<u>Elite</u> NON-INJ: 181.0 ± 5.0 INJ: 178.0 ± 4.0 <u>Sub-elite</u> NON-INJ: 176.0 ± 7.0 INJ: 177.0 ± 5.0
Pruyn et al. 2012	Australian Rules Football Players	NON-INJ: n = 11 INJ: n = 28	NON-INJ: 23.5 ± 4.6 INJ: 24.2 ± 4.2	NON-INJ: 84.4 ± 5.2 INJ: 88.4 ± 9.1	NON-INJ: 185.0 ± 8.0 INJ: 188.0 ± 7.0
Pruyn et al. 2013	Australian Rules Football Players	n = 25	24.9 ± 4.3	86.8 ± 8.1	187.0 ± 7.3
Serpell, Scarvell et al. 2014	Australian Rules Football Players	NON-INJ: n = 18 INJ: n = 13	NON-INJ: 22.7 ± 3.1 INJ: 20.5 ± 2.1	NON-INJ: 83.1 ± 6.7 INJ: 85.4 ± 9.8	NON-INJ: 184.8 ± 8.2 INJ: 189.8 ± 8.3
Watsford et al. 2010	Australian Rules Football Players	NON-INJ: n = 122 INJ: n = 14	NON-INJ: 22.6 ± 3.5 INJ: 27.0 ± 3.4	NON-INJ: 86.4 ± 8.5 INJ: 87.1 ± 8.4	NON-INJ: 187.4 ± 7.6 INJ: 186.6 ± 7.6

Note. NON-INJ Non-injured, INJ Injured

3.4.4 Stiffness Assessment

A breakdown of how stiffness was assessed in the included studies is reported in Table 3.2. All six studies utilised a force plate/platform for stiffness assessment, with one also employing the use of a high-speed camera in order to digitise the movement (Nagahara & Zushi, 2017). This same study was also the only study to determine stiffness through two maximal effort 60m sprints (Nagahara & Zushi, 2017). However, only the left limb was assessed. Four of the six studies conducted a unilateral hopping test in which three consecutive hops were analysed to represent leg stiffness across a trial (Pickering Rodriguez et al., 2017; Pruyn et al., 2012, 2013; Watsford et al., 2010). Frequency was set at 2.2Hz for each of these studies as it is the preferred frequency for human hopping (Farley et al., 1991). It is also representative of the lowest frequency in which the leg behaves like a spring (Farley et al., 1991). The remaining study asked participants to perform a double leg rebound jump at a self-selected pace (Serpell, Scarvell, et al., 2014). Only the second jump was analysed for stiffness assessment. These five studies assessed both left and right limbs.

The way in which stiffness was calculated varied between the six studies. Three of the studies calculated leg/vertical stiffness as PVGRF divided by COM during the middle of ground contact (Pickering Rodriguez et al., 2017; Pruyn et al., 2012, 2013). This method has been previously validated by Ferris and Farley (1997); McLachlan et al. (2006). One study calculated leg stiffness through specialised software using the methods explored by Farley et al. (1991); Farley and Morgenroth (1999) (Watsford et al., 2010). The other study examining stiffness through simple jump tasks calculated vertical stiffness as ground reaction force divided by COM displacement (Serpell, Scarvell, et al., 2014). However, there was no reference stated to determine where this stiffness calculation was from. The only study to determine stiffness through sprinting utilised the linear approximation of ground reaction forces in relation to downward centre of gravity displacement for vertical stiffness, with respect to the shortening quasi length for leg stiffness and the linear approximation of net joint moment corresponding to the decrease in joint angle during the support phase for joint stiffness (Nagahara & Zushi, 2017). The kinematics of the segments and joints were calculated from the smoothed coordinate data.

The frequency of stiffness assessment and duration of each study also varied. Two studies assessed stiffness at least once per month, up to once per week depending on athlete availability to form a monthly leg stiffness score (Pruyn et al., 2012, 2013). Stiffness was assessed one month prior to and continued throughout the regular 2009 AFL season. One study only assessed leg stiffness one month prior to the commencement of the 2006 AFL season (Watsford et al., 2010). These leg stiffness values were compared to players who suffered a non-contact soft tissue hamstring injury throughout the season. One study assessed stiffness before and after the completion of a six-month winter training period (Nagahara & Zushi, 2017). Another study assessed vertical stiffness once during pre-season and once per month for the duration of the 2013 domestic netball season (Pickering Rodriguez et al., 2017). The other remaining study assessed vertical stiffness on a weekly basis, 72 hours after match play (Serpell, Scarvell, et al., 2014). However only three-time points were reported in the study: end of preseason, near to 3 weeks prior to injury occurrence and within one week of injury occurrence. The total duration of data collection was over two competitive football seasons. It is unclear as to which football season these time points are from or if data from the two seasons were combined. Information on reported statistics, results and conclusions can be found in Table 3.3.

Table 3.2 *Stiffness Assessment*

Authors	Equipment	Task	Stiffness Measure	Frequency of Stiffness Assessment	Duration of Study
Nagahara et al. 2017	Laser (50Hz, LAVEG-Sport Jenoptic, Jena, Germany) High speed video camera (HSV-500C, Nac, Kanagawa, Japan) Two force platforms (0.6 x 1.2m, Kistler, Winterthur, Switzerland)	60m sprint. Left foot contact analysed	$K_{vert,}$ K_{leg} K_{joint}	Once pre-season (October), Once post season (April)	Six – month winter training period
Pickering Rodriguez et al. 2017	Force platform (Onspot, Wollongong, Australia)	Unilateral hopping task. 3 consecutive hops (frequency 2.2Hz) analysed.	K_{vert}	Once per month	Once during pre-season and throughout the 2013 season
Pruyn et al. 2012	One dimensional force platform (Onspot, Wollongong, NSW)	Unilateral hopping task. 3 consecutive hops (frequency 2.2Hz) analysed.	K_{leg}	Once per month, up to once per week	One month prior and throughout the 2009 AFL regular season.
Pruyn et al. 2013	One dimensional force platform (Onspot, Wollongong, Australia)	Unilateral hopping task. 3 consecutive hops (frequency 2.2Hz) analysed.	K_{leg}	Once per month, up to once per week	One month prior and throughout the 2009 AFL regular season
Serpell, Scarvell et al. 2014	Fitness Technology 400 series force plate (Fitness Technology, Adelaide, Australia)	Double leg rebound jump (self-selected pace). Second jump analysed.	K_{vert}	Weekly basis 72rs after playing a game	Two competitive football seasons
Watsford et al. 2010	Force platform (Onspot, Wollongong, New South Wales, Australia)	Unilateral hopping task. 3 consecutive hops (frequency 2.2Hz) analysed.	K_{leg}	One month prior to the commencement of the competitive season	2006 AFL season.

Note. K_{leg} Leg Stiffness; K_{vert} Vertical Stiffness; K_{joint} Joint Stiffness; AFL Australian Football League

Table 3.3 *Statistics, key findings, conclusions*

Authors	Statistics	Key Findings	Conclusion
Nagahara et al. 2017	Paired t-test, Cohen's *d*, Pearson's product moment correlation coefficient	Between pre-test and post-test: ▪ K_{vert} significantly increased (*p*=0.001, ES=1.53) ▪ No significant difference for K_{leg} (*p*=0.686, ES=0.12) ▪ Ankle K_{joint} significantly increased (*p*=0.002, ES=1.41) ▪ No significant difference for K_{joint} (*p*=0.448, ES=0.18) ▪ No significant relationships were found between running speed and changes in K_{vert}, K_{leg}, knee K_{joint} and ankle K_{joint}	Developed maximal speed sprinting performance may be accompanied by an increase in K_{vert} and ankle K_{joint}.
Pickering Rodriguez et al. 2017	Student's t-test, Mann-Whitney U test	▪ No difference between mean K_{vert} scores between INJ and NON-INJ groups for: - Entire cohort (*p*=0.556, *d*=0.22) - Elite group (*p*=0.418, *d*=0.58) - Sub-elite group (*p*=0.820, *d*=0.11) ▪ Injury segmentation analysis revealed no significant difference in K_{vert} for: - Entire cohort (OR: 1.00, RR: 1.00) - Elite group (OR: 0.44, RR: 0.67) - Sub-elite group (OR: 1.75, RR 1.50)	Increased incidence of injury is related to greater levels of stiffness in the Achilles and Soleus.
Pruyn et al. 2012	Independent samples t-test, one-way ANOVA	▪ No difference between mean K_{leg} scores for NON-INJ and INJ group (*p*=0.721). ▪ INJ group significantly higher bilateral differences in K_{leg} than NON-INJ group (*p*=0.05). ▪ Players who suffered an injury as a result of running/sprinting or twisting/bending were found to have significantly higher bilateral differences across the season compared with the NON-INJ group (*p*=0.004 respectively).	Players with a moderately high bilateral difference in K_{leg} may be more prone to non-contact, soft tissue injuries in the lower body.

Pruyn et al. 2013	One-way ANOVA, Tukey's post hoc test	▪ No significant differences between mean K_{leg} scores (p=0.32) and mean bilateral differences (p=0.774) from each month.	K_{leg} does not significantly change throughout a season of professional AFL football.
Serpell, Scarvell et al. 2014	Independent samples t-test, 2-way ANOVA, Pearson's Correlation	▪ No significant differences in K_{vert} means were found between groups (absolute p=0.18, relative p=0.08) or within groups (absolute p=0.83, relative p=0.88). ▪ No relationship was evident between K_{vert} and age (absolute r=-0.06, relative r=-0.06). ▪ No relationship was evident between K_{vert} and training history (absolute r=-0.01, relative r=0.00).	K_{vert} is not related to lower limb muscle strain injury in Australian Rules footballers.
Watsford et al. 2010	Independent samples t-test, Paired t-tests	▪ Players who sustained an acute hamstring injury were significantly older (p<0.01) and recorded significantly higher bilateral mean K_{leg} values (p=0.03). ▪ INJ players: K_{leg} of the involved limb was significantly higher than the NON-INJ group mean (p=0.02). ▪ No differences between involved and non-involved limb for INJ group (p=0.58).	Higher bilateral K_{leg} may be a factor in sustaining a non-contact, soft tissue hamstring injury.

Note. INJ Injured, NON-INJ Non-injured, K_{leg} Leg stiffness, K_{vert} vertical stiffness, K_{joint} Joint stiffness, ANOVA analysis of variance, OR odds ratio, RR relative risk, ES effect size

3.4.5 Study Quality Assessment

Studies selected for inclusion were assessed for quality based on the Strengthening the Reporting of Observation Studies in Epidemiology (STROBE) criteria (von Elm et al., 2007) (Appendix A). Although the included studies were not specifically epidemiologically focused, it was deemed appropriate to implement the STROBE criteria as studies were fundamentally based on a research design similar to epidemiological studies. The STROBE criterion gives clarity to the quality of reporting rather than quality of investigations, design and/or methods. The STROBE statement consists of 22 items. A breakdown of how each criterion related to each of the studies included for review can be seen in Table 3.4.

Table 3.4 *STROBE criterion breakdown of articles included for review*

		Nagahara et al. 2017	Pickering et al. 2017	Pruyn et al. 2012	Pruyn et al. 2013	Serpell, Scarvell et al. 2014	Watsford et al. 2010
Title and Abstract	1 (a)	×	×	×	✓	×	✓
	1 (b)	✓	✓	✓	✓	✓	✓
Introduction	2	✓	✓	✓	✓	✓	✓
	3	×	✓	✓	✓	×	✓
Methods	4	×	×	×	✓	✓	×
	5	✓	✓	✓	✓	✓	✓
	6 (a)	NA	×	×	×	×	×
	6 (b)	NA	NA	NA	NA	NA	NA
	7	×	✓	×	✓	×	×
	8	✓	✓	✓	✓	✓	✓
	9	×	×	×	×	×	×
	10	×	×	×	×	×	×
	11	×	×	×	×	×	×
	12 (a)	✓	✓	✓	✓	✓	✓
	12 (b)	NA	✓	✓	NA	✓	✓
	12 (c)	NA	NA	NA	NA	✓	NA
	12 (d)	NA	NA	NA	NA	NA	NA
	12 (e)	NA	NA	NA	NA	NA	NA
Results	13 (a)	✓	✓	✓	✓	✓	✓
	13 (b)	NA	NA	NA	NA	NA	NA
	13 (c)	NA	NA	NA	NA	NA	NA
	14 (a)	✓	✓	✓	✓	✓	✓
	14 (b)	NA	NA	NA	NA	NA	NA
	14 (c)	NA	✓	✓	✓	✓	NA
	15	×	×	×	✓	✓	✓
	16 (a)	×	✓	×	×	×	×
	16 (b)	NA	NA	NA	NA	NA	NA
	16 (c)	NA	NA	NA	NA	NA	NA
	17	NA	×	×	NA	×	×
Discussion	18	✓	✓	✓	✓	✓	✓
	19	×	✓	✓	✓	✓	✓
	20	✓	✓	✓	✓	×	✓
	21	×	×	×	×	×	×
Other Information	22	×	×	×	✓	×	✓

✓ Meets criteria; × Does not meet criteria; NA not applicable

3.5 DISCUSSION

The primary objective of this review was to evaluate the populations, methodologies and findings used in the current literature surrounding the longitudinal assessment of lower limb stiffness in athletic populations. With only six studies included for review, it is evident that limited longitudinal studies examining lower limb stiffness and its links with performance and injury exist.

This systematic review combined six studies with a total of 269 participants. Of the studies included for review, four of them recruited professional Australian Rules football players, one recruited nine well-trained male track and field athletes, while the other study recruited a combination of elite and sub-elite netballers. Due to the specific nature of recruitment, it is evident that the application of these findings to other sports may be limited. However, it is acknowledged that the recruitment of elite sporting populations in research is often very challenging.

In the examined literature, simple vertical hopping and jump tasks were the predominant tasks used to assess lower limb stiffness. A recent study by Maloney et al. (2016) highlighted the difference between such tasks. Despite no significant difference, greater vertical stiffness asymmetries were found in acyclic activities (i.e. bilateral and unilateral drop jumps) when compared to a cyclic submaximal task (i.e. bilateral hopping) (Maloney et al., 2016). Although the reliability of drop jump derived stiffness measures are lacking within the current literature (Maloney, Richards, & Fletcher, 2018), it is not surprising that greater vertical asymmetries were found in the drop jump tasks. Instinctively, it appears that by increasing drop jump intensity (i.e. height of box and subsequent vertical ground reaction force), greater vertical stiffness asymmetries would be evident even though further investigation is needed (Maloney et al., 2016). While it may appear that acyclic activities are better predictors of vertical stiffness asymmetries, sport practitioners should carefully consider how the limbs function during sport performance as stiffness has been found to be dependent on task demands (Komi, 2000; Kuitunen et al., 2002). The varying nature of sporting demands and the way an athlete deals with mechanical load is thought to influence performance and injury risk. Due to the lack of longitudinal tracking and regular stiffness assessment using sport specific measures, prospective links between stiffness, performance and injury remain unclear. However, it is thought that the use of sport specific measures may provide a clearer insight into such links.

Stiffness assessment has traditionally involved the use of a force plate. All the studies included for review utilised a force plate for stiffness assessment. Although a valid and reliable measure, force plates are limited in size, often fixed in position (i.e. embedded in the ground especially for running trials) and thereby limit the amount of data that can be obtained (Brughelli & Cronin, 2008b; Hood et al., 2012; Raymond et al., 2018). For example, Nagahara and Zushi (2017) reported that when an athlete missed the force plate with their whole left foot the sprint had to be repeated after a rest period of fifteen minutes. The ability to use this equipment in the field is limited hence why lower limb stiffness assessment has been isolated to laboratory-based settings and has not been undertaken on a regular basis. It is important for researchers to consider field-based options provided they have been appropriately validated (i.e. contact mat, phone applications) (Balsalobre-Fernández et al., 2017; Dalleau et al., 2004) as this will make the utilisation for daily/weekly/monthly monitoring more accessible.

Of the studies included for review, only one attempted to assess the stiffness of all participants on a weekly basis (i.e. 72 hours after match play) (Serpell, Scarvell, et al., 2014). Despite this, the study failed to report and examine stiffness changes across two football seasons instead opting to only include three-time points of which no relationship between vertical stiffness and injury was found. The use of a single jump for stiffness assessment places questions around the accuracy of data obtained as a study assessing data reduction on inter-trial variability of lower limb stiffness suggested that one contact is insufficient with at least three consecutive contacts considered optimal (Moresi et al., 2015). It remains unknown to what extent a single jump accurately reflects an individual's level of stiffness given that repeat jump tasks represent common activities used to assess lower limb stiffness within the current research literature (Butler et al., 2003). Additionally, the incorporation of three-time points across two football seasons does not adequately reflect long-term stiffness changes or how stiffness varied on a weekly basis which may provide a valid reason as to why no relationship between vertical stiffness and lower limb muscle strain injuries were found in Australian Rules footballers.

Two of the studies assessed leg stiffness once per month, up to once per week depending on the availability of participants (Pruyn et al., 2012, 2013). Despite not all participants being tested as frequently as weekly but no less than monthly, one study found a significant relationship between mean bilateral differences in leg stiffness, with the injured group reporting a much higher difference than the non-injured group (Pruyn et al., 2012). No difference between mean leg stiffness measures were found between groups (Pruyn et al., 2012). The other remaining study examining AFL players assessed leg stiffness one month prior to the commencement of a football season (Watsford et al., 2010). Leg stiffness values were compared to players who suffered a non-contact soft tissue hamstring injury throughout the season in order to identify whether stiffness was related to injury occurrence. It appears higher bilateral leg stiffness may be a factor in sustaining a non-contact, soft tissue hamstring injury. However, as noted by the authors an important consideration of this study is that these findings may not reflect the stiffness measure at the time of injury given that leg stiffness was assessed in the preseason and injuries occurred at different stages throughout the season. In addition, it is important to note that only simple between-group analyses were conducted in these two studies, which may have neglected important information regarding an individual's change to stiffness.

Four of the six studies included for review compared lower limb stiffness between an injured and non-injured group (Pickering Rodriguez et al., 2017; Pruyn et al., 2012; Serpell, Scarvell, et al., 2014; Watsford et al., 2010). Although similar, these studies adopted varying definitions of injury. However, with the exception of the sub-elite group in the study by Pickering Rodriguez et al. (2017), all of the studies incorporated trained clinicians to diagnose injury, which is in advancement of self-report measures used in other literature.

The only study examining netball and a female population reported no significant difference in vertical stiffness between the injured and non-injured groups during repeat hops (Pickering Rodriguez et al., 2017), which contrasts the earlier findings of Watsford et al. (2010). Although lower limb stiffness has been found to vary between genders (Granata et al., 2002), it is unknown to what extent stiffness is affected by different training modalities, match play and playing position (Pickering Rodriguez et al., 2017). Of the studies included for review, only one accounted for training load (Pruyn et al., 2013). Although no relationship was found between leg stiffness and training load, the authors revealed that a substantial variation in training load across the football season was evident. As identified by the authors a possible explanation as to why leg stiffness did not vary across a football season may be a result of considering stiffness scores as a group mean thereby nullifying any individual differences that may have occurred. For coaches, the value of monitoring training loads in athletes comes with tracking individual rather than team scores. Further research exploring such a relationship is needed to maximise performance and provide a greater understanding of an individual's tolerance to training load, which will hopefully minimise injury risk.

In addition, the only study assessing stiffness through maximal sprinting also considered differences as a group mean (Nagahara & Zushi, 2017). The study assessed stiffness prior to and following the completion of a six-month training period. However, the researchers did not implement a specific training program for athletes to follow given their different event backgrounds. The researchers also only examined left foot contacts. Despite this, a significant increase in vertical and ankle joint stiffness was reported between pre and post testing. Though it is unclear as to whether a certain athletes training background/program contributed to these findings or if any athletes were injured throughout the six-month duration.

The relevance of existing knowledge and its practical application for coaches and athletes is limited due to the lack of information based on the assessment of sport specific tasks and evaluation of diverse athletic populations. The only research assessing longitudinal assessment of lower limb stiffness in athletic populations has focused on AFL, elite and sub-elite netball players during simple vertical hopping and jump tasks and a small number of track and field athletes during maximal sprinting. Stiffness assessment hasn't been reported to occur as frequently as daily/weekly, which may offer an explanation as to why limited long-term stiffness changes have been found. This is a possible limitation of the literature surrounding longitudinal stiffness assessment as a change may occur and be subsequently missed as a result of infrequent measures. It is important that researchers consider the availability of field-based equipment to assess lower limb stiffness as this may increase the occurrence of lower limb stiffness monitoring in athletic populations. By doing so, it may detail the way an athlete deals with load which could provide vital feedback to athletes, coaches, medical and support staff enabling early injury risk identification and performance maximisation.

The future of monitoring an athlete's response to load will more than likely be dominated by the emergence of new technologies (Foster, Rodriguez-Marroyo, & de Koning, 2017). Currently, sport practitioners don't have adequate evidence to inform coaches about daily changes in an athletes training status and how this influences performance (Foster et al., 2017). If accurate and reliable information can be provided to coaches in real-time, it may enhance their knowledge of individual training responses, assist in the development/adaptation of training programs and most importantly provide an additional avenue for communication between sport practitioners, coaches and athletes ultimately enhancing performance. However, given the realities of new technologies and their ability to provide so much information (Foster et al., 2017), it is important that sport practitioners don’t overload coaches as this may become detrimental to an athletes performance.

3.6 CONCLUSION

It is evident from this systematic review that limited longitudinal studies examining lower limb stiffness exist despite suggested links with performance and injury. The majority of studies focused on the specific sporting population of AFL players and mainly assessed stiffness through simple vertical hopping and jump tasks on a force plate. There is a need to examine longitudinal stiffness changes in varying athletic populations using more functional or sport specific lower limb stiffness measures at more regular intervals to further investigate the prospective links between lower limb stiffness, performance and injury. With the emergence of new technologies, this may now be a possibility. Not only are these technologies thought to increase the occurrence of lower limb stiffness monitoring but also detail regular changes relevant to an athletes training environment.

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CHAPTER 4: EXTENDED METHODOLOGY

This chapter provides an overview of the generic methodological elements related to the entire research project (i.e. research design, participant recruitment and general testing procedures).

4.1 RESEARCH DESIGN

This research employed a cross-sectional study design investigating high-level track and field athletes to validate the use of inertial measurement units in obtaining a lower limb stiffness running measure.

4.2 INERTIAL MEASUREMENT UNIT

The inertial measurement units used for this research project are known as the IMeasureU sensors. The IMeasureU sensors (I Measure U, Auckland, New Zealand) are lightweight versatile inertial measurement units that record precise movements in 9-axis, where measurements can either be stored on board, or transmitted wirelessly to a smart phone in real time (Figure 4.1). The sampling frequency can range between 100 and 1000Hz. The motion is captured with three different internal sensors:

- Accelerometer: 3-axis range of ±16g; at a 16bit resolution
- Gyroscope: 3-axis full scale range of ±2000°/sec; at a 16bit resolution
- Magnetometer: 3-axis full scale range of ±1200mT, at a 13bit resolution

Figure 4.1 IMeasureU sensor

4.3 ETHICS APPROVAL

Ethical approval was granted by the Australian Catholic University Human Research Ethics Committee (Approval no. 2016-284H) (Appendix B). Following ethical clearance, all participants were provided with a summary of the proposed research (Appendix C) along with an informed consent form (Appendix D). Athletes under the age of eighteen were provided with an additional consent form (Appendix D), as they required parent/guardian approval.

4.4 PARTICIPANTS

4.4.1 Population

Nineteen high-level track and field athletes (7 male, 12 female; Age: 22.05 ± 3.39years, Height: 1.71 ± 0.09m, Weight: 62.56 ± 11.66kg) were recruited to participate in this study. These nineteen athletes were made up of six sprinters, seven middle distance runners and six hurdlers (Table 4.1).

Table 4.1 *Descriptive characteristics of athletes by event group*

	Sprinters	Middle Distance	Hurdlers
Age (yrs)	22.50 ± 4.32	22.14 ± 3.85	21.50 ± 2.07
Height (m)	1.70 ± 0.09	1.68 ± 0.09	1.75 ± 0.10
Weight (kg)	64.70 ± 9.91	54.99 ± 8.70	69.25 ± 12.69
Training Hours p/w	11.17 ± 3.19	12.29 ± 6.10	14.33 ± 2.25

Note. p/w per week

4.4.2 Inclusion and exclusion criteria

An inclusion and exclusion criteria were established in order to reduce participant variability and ensure that the population was representative of the groups in which this research targeted. Participants were high-level, able-bodied athletes defined by Athletics Australia qualifying standards for relevant national team representation (i.e. Oceania Championships, World Junior Championships, World Youth Championships, Commonwealth Games, Olympic Games, World Championships and World University Games). At the time of testing, all participants needed to be free from a lower body injury and be completing full training.

4.4.3 Screening (Questionnaire)

In order to gather essential descriptive data related to stiffness measures, participants were screened for age, athletic background, highest level of competition, training years, leg dominance, exercise and injury history (Appendix E).

4.5 DATA COLLECTION

4.5.1 Anthropometric Measures

Basic anthropometric measures including height, weight, limb lengths and joint widths were measured to provide descriptive data as well as for input into the motion analysis model prior to the commencement of testing procedures (Table 4.2). Height (cm) and weight (kg) were measured using a stadiometer, accurate to ± 0.01 m (SECA height rod model, Hamburg Germany) and digital scales (Tanita BWB-600 Digital Medical Scales, Wedderburn, Australia) accurate to ± 0.05kg. Limb lengths and joint widths were measured using an anthropometer (accurate to 0.001m). All anthropometric measures were taken using methods of the international Society for the Advancement of Kinanthropometry (ISAK).

Table 4.2 *Required limb lengths and joint widths for PluginGait (VICON; Oxford Metrics Ltd, Oxford, United Kingdom)*

Measure	Description
Leg Length	Measured from the Anterior superior iliac spine to medial malleolus while participant is standing.
Knee Width	Medio-lateral width of knee across the line of the knee axis
Ankle Width	Medio-lateral distance across the malleoli
Elbow Width	Width of the elbow along flexion axis between the medial and lateral epicondyles of the humerus
Wrist Width	Anterior/posterior thickness of the wrist as position where wrist marker is attached
Hand Thickness	Anterior/posterior thickness between the dorsal and palmar surfaces of the hand
Shoulder Offset	Vertical offset from shoulder joint centre to base of acromion marker

4.5.2 Plug-in Gait model

The standard 39 marker PluginGait Fullbody model (Vicon, Oxford Metrics Ltd, Oxford, United Kingdom) was utilised to obtain kinematic data with reflective markers attached to appropriate body landmarks with double sided tape and secured with hypoallergenic medical tape (Fixomull, Smith & Nephew, London, UK) (Figure 4.2).

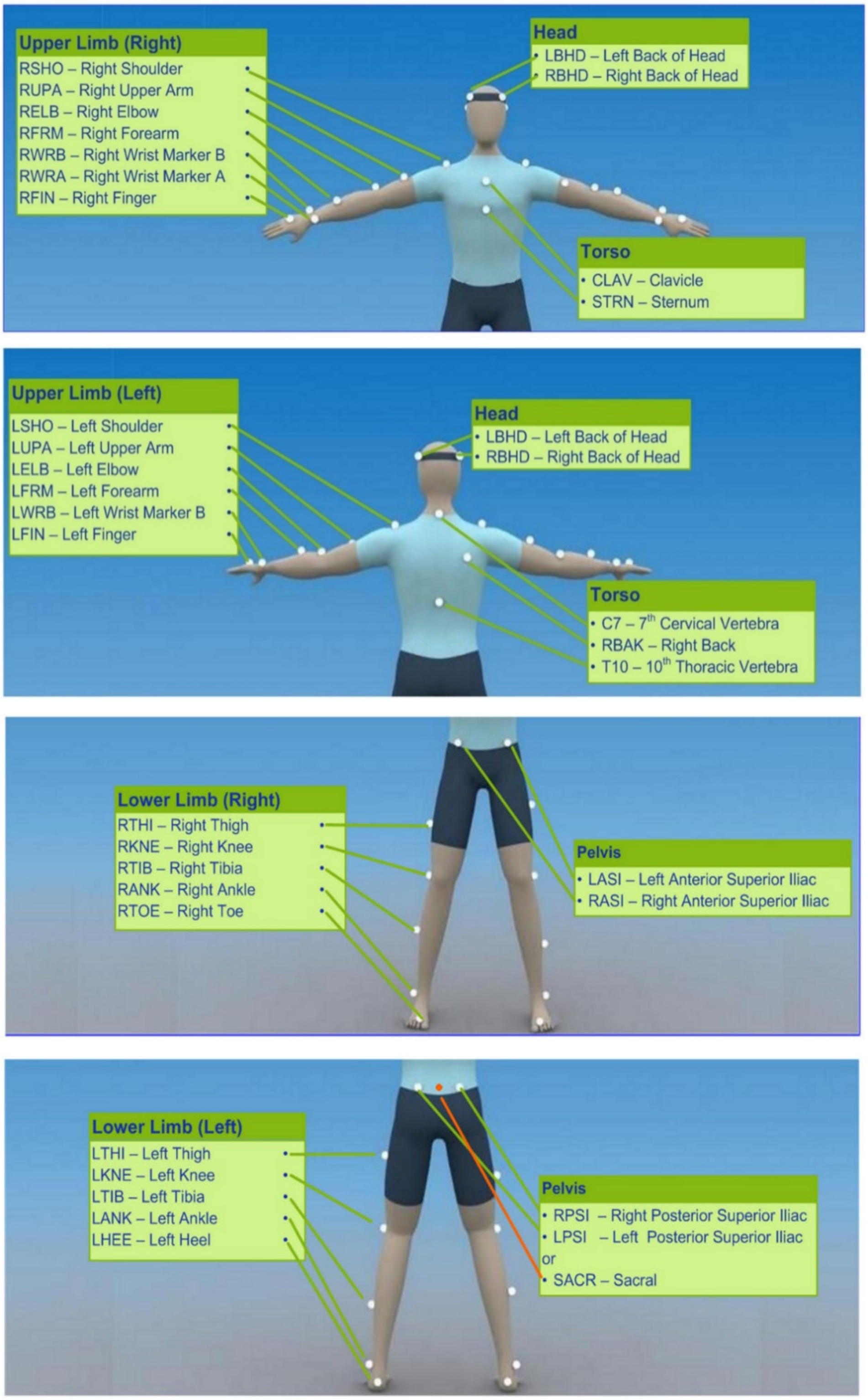

Figure 4.2 *Vicon PluginGait Fullbody marker placements (Vicon; Oxford Metrics Ltd, Oxford, United Kingdom)*

4.5.3 Testing Procedures

Following a self-directed warm up (approximately 30 minutes), which consisted of a full body activity (i.e. jogging, cycling), dynamic stretching and event specific drills, participants were instructed to perform six running gait trials from a standing start at a pace reflective of their event during competition. There were no restrictions placed on the participants starting position. However, running speeds were monitored using timing gates placed 20m apart in the testing area to ensure consistency between trials. The distance of every run was also measured to input into the MATLAB algorithm. The distances run ranged between 40 and 51m. Along with ensuring consistent running speeds, participants needed to attain six-foot strikes on the force plate (i.e. three left foot contacts and three right foot contacts). If the participant missed the force plate, the trial had to be repeated to ensure that all the necessary foot contacts were attained. Participants were required to wear tight bike shorts, crop top (females) and running spikes to allow adequate exposure of all relevant markers and IMeasureU sensor placement. Running spikes were worn to mimic an athlete's training and competition environment.

All testing was performed on an indoor 55m Mondo running track in the biomechanics laboratory at the New South Wales Institute of Sport. A fourteen camera, 3D motion analysis system (VICON Vantage 5; Oxford Metrics Ltd, Oxford, United Kingdom) sampling at a rate of 250Hz was used to determine full body kinematic data during each trial. Four floor mounted force plates (Kistler 9281CA, Winterthur, Switzerland) (situated 30m along the 55m running track) sampling at a rate of 1000Hz were used simultaneously to measure ground reaction forces during each trial (Figure 4.3). A handheld Panasonic camera (positioned behind the participant) and two Vicon Vue cameras (placed side and front on in the capture zone) were used to capture high speed video for participant feedback as well as to determine when and where each foot strike occurred. Adequate recovery time was provided between each trial in order to avoid the influence of fatigue on the required stiffness measures.

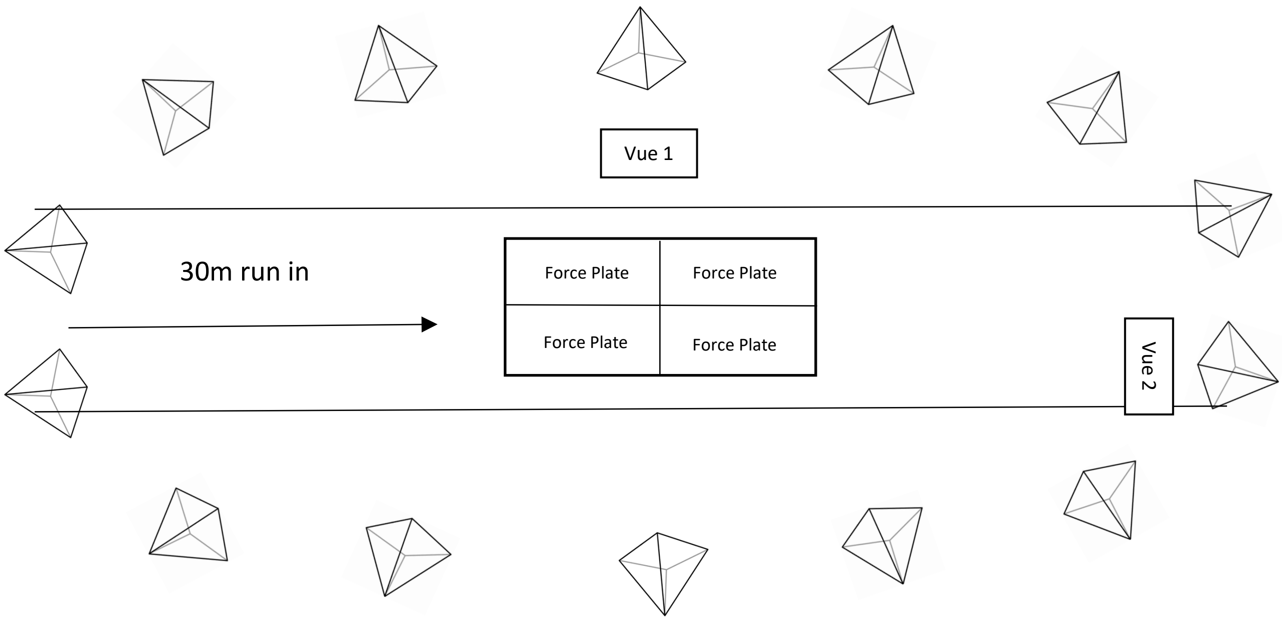

Figure 4.3 *Laboratory configuration*

Three IMeasureU sensors (I Measure U, Auckland, New Zealand) (Firmware version 1.3.0) sampling at 500Hz were placed on both the left and right distal tibias as well as in the middle of the left and right posterior superior iliac spine (PSIS) markers for each participant (i.e. approximate COM location). The IMeasureU sensor placed on the mid PSIS was secured with medical adhesive tape while a Velcro strap attached both tibia sensors. (Figure 4.4). All IMeasureU sensors were placed on the participant following marker placement and were attached without discomfort or in a way that restricted movement but ensured minimal artifactual movement of the sensor. It should be noted that the IMeasureU sensors were not synced with Vicon due to the IMeasureU-Vicon sync bandwidth approach (i.e. the accelerometer, magnetometer and gyroscope data (9axis) can only be captured with a sample frequency of 100Hz).

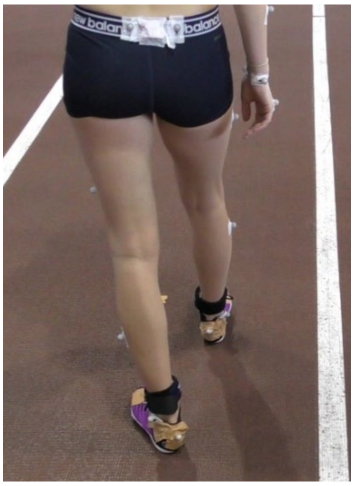

Figure 4.4 IMeasureU sensor placement

All three IMeasureU sensors were controlled by the IMeasureU Research Application (App) (Version 3.3) on an iPhone using the on-board logging method. The on-board method stores data on the sensor's internal memory and allows sampling at a higher frequency. It also allows the sensor to disconnect from the App and continue to freely record data. Figure 4.5 shows the initial setup of the sensors in the App. Once the sensors were configured, the sensors had begun logging and the participants were free to walk away from the phone.

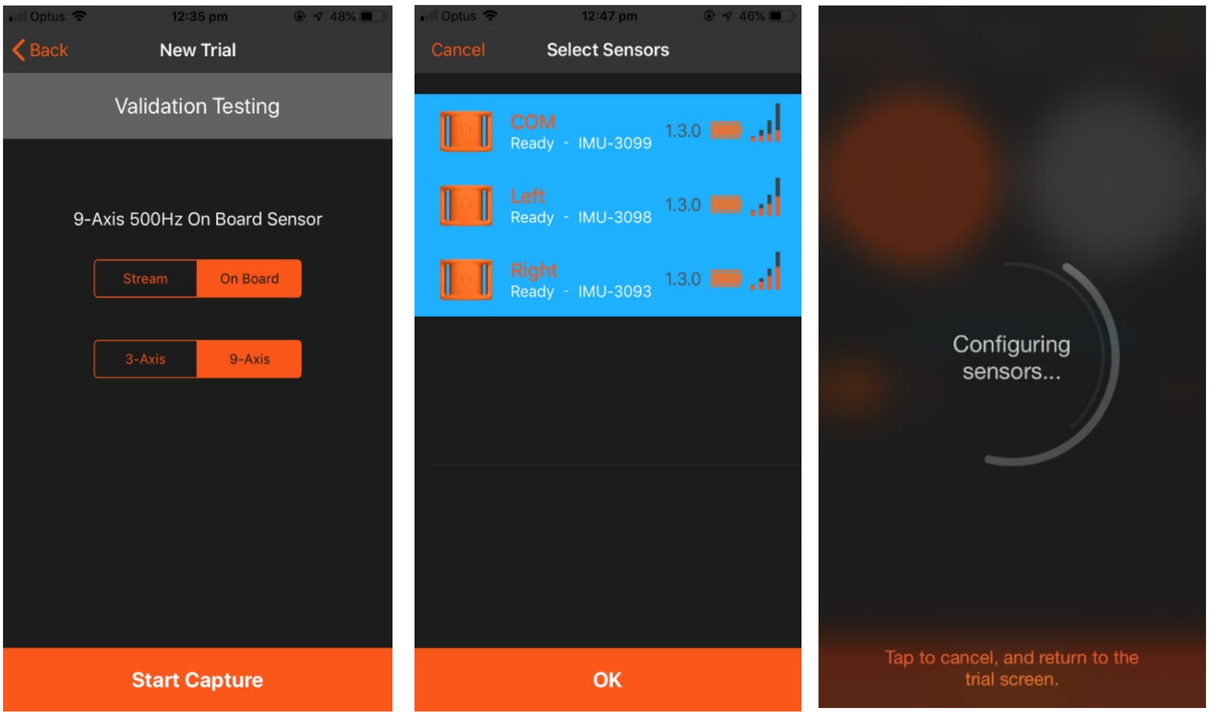

Figure 4.5 Initial setup of sensors in IMeasureU Research App

The IMeasureU sensors were logged for the whole testing duration. However, the start and stop functions on the App (Figure 4.6) were used to create timing periods for each trial in the synchronisation data.

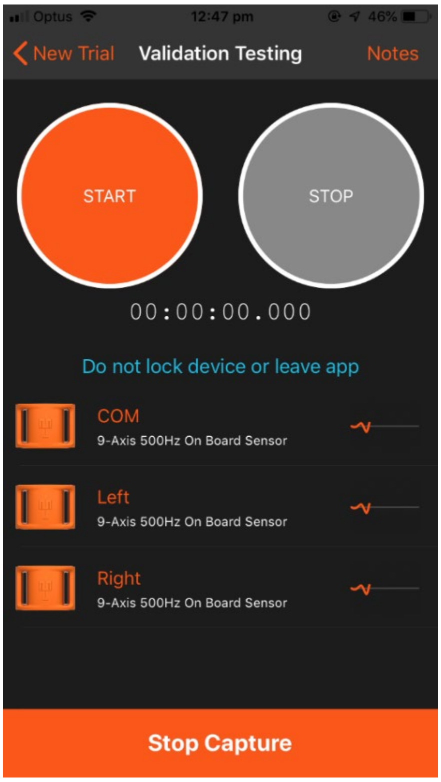

Figure 4.6 Start/stop buttons to create timing periods in IMeasureU Research App

Once all the trials were completed, the IMeasureU sensor capture was stopped (Figure 4.7) with the .csv files for each individual sensor exported from the App via email (Figure 4.8).

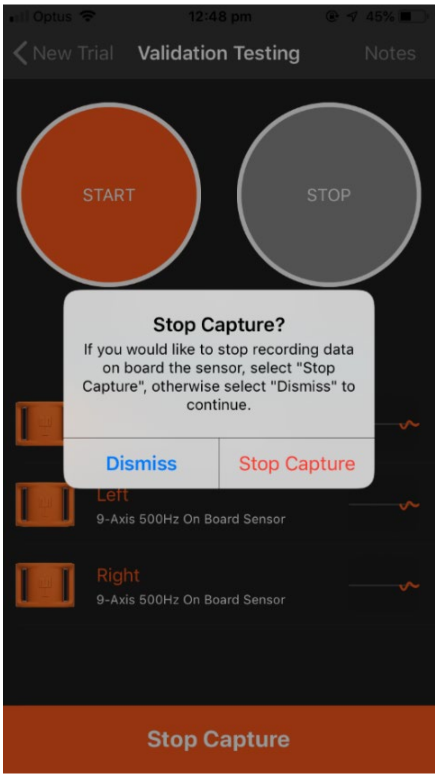

Figure 4.7 Stop capture of data in IMeasureU Research App

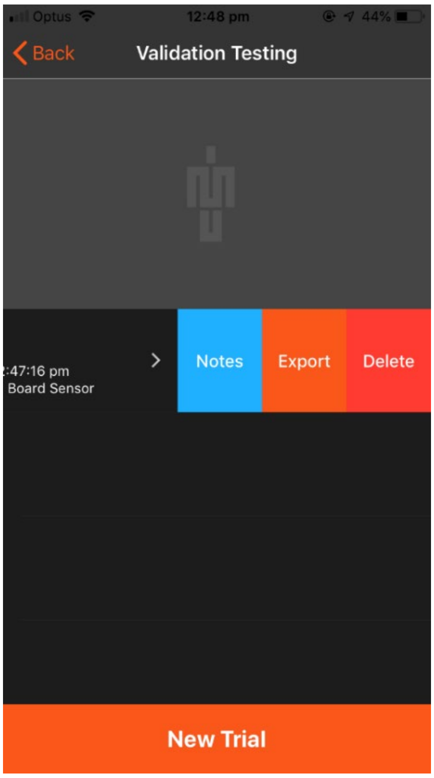

Figure 4.8 Exporting .csv files from IMeasureU Research App

Each IMeasureU sensor was then plugged into the computer and the .csv files were synced with the data stored onboard each unit through the Lightning Desktop App (Version 3.0.0).

4.6 DATA ANALYSIS

4.6.1 Vicon Data

A dual low pass, fourth order Butterworth filter with a cut off frequency of 23Hz was used to smooth the kinematic data (Millett, Moresi, Watsford, Taylor, & Greene, 2014). Following filtering, the PluginGait Fullbody model was used to find PVGRF and calculate COM displacement. Vertical stiffness was calculated as PVGRF divided by maximum vertical COM displacement and leg stiffness was calculated as the maximum vertical force divided by the change in vertical leg length (McMahon & Cheng, 1990). Only the eccentric displacement during landing was used to calculate stiffness.

4.6.2 IMeasureU Data

The raw IMeasureU data for each participant were imported into a custom-made MATLAB algorithm (R2016b, The Mathworks Inc, Natick, MA), in which all three IMeasureU sensors were synced together. An explanation of how the algorithm worked is described below.

Firstly, the algorithm required the input of the number of tibia sensors that were worn during the data collection. For our data collection, this number was always 2. The COM file was then inputted into the algorithm for one participant, followed by the left and right leg sensor files for the same participant. The algorithm then displayed an image of all the trials captured for that participant (Figure 4.9).

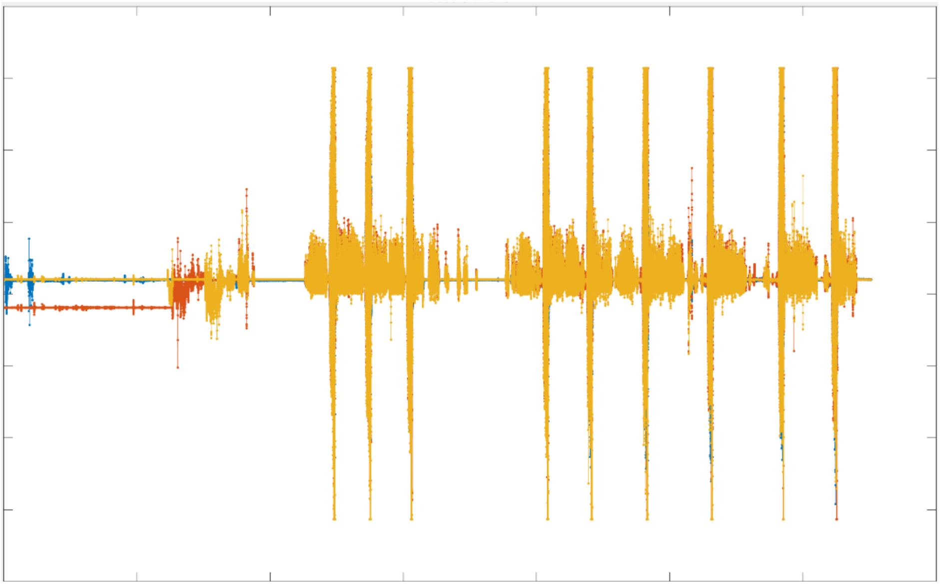

Figure 4.9 Synced data of all trials for one participant in MATLAB

We then were required to select the correct sprint we wanted to analyse. Once the sprint we wanted to analyse had been selected, the resultant accelerations for both limbs were calculated, and the cropped data was plotted (Figure 4.10).

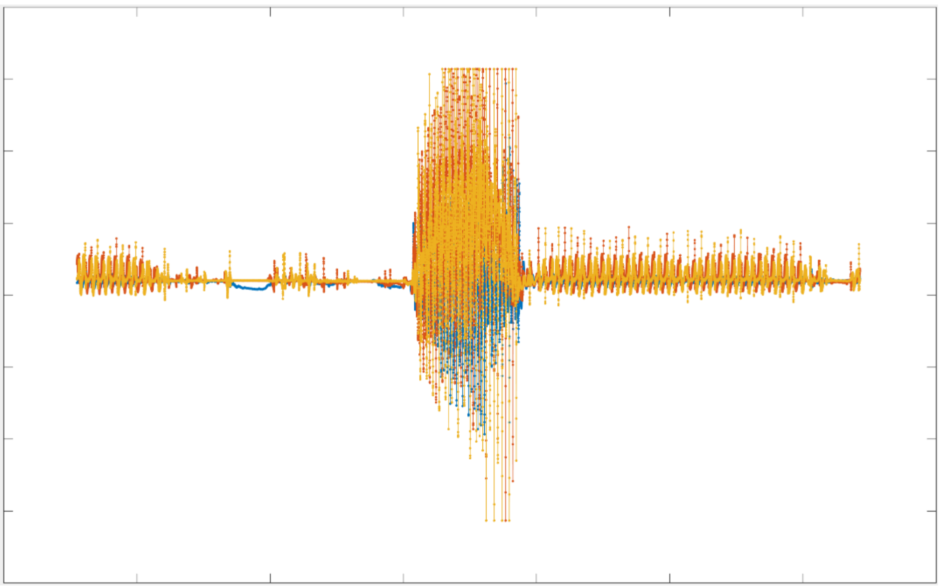

Figure 4.10 *Cropped data to show selection of one trial in MATLAB*

The algorithm was configured to receive and process the raw 3 axis acceleration data and calculate the resultant acceleration magnitude data at each time sample. The peak shock for each individual foot strike were determined by the peak magnitude/highest value of the resultant acceleration vector at the location of the heel strike in each foot strike. This was calculated by square rooting the sum of the squares for all three acceleration measures at each time point. The resultant magnitude data was then filtered using a bandpass filter to filter out high frequency noise (i.e. skin movement) and very low frequency movement. The filtered resultant acceleration data was then processed to identify the fundamental frequency. A fast Fourier Transform was used to find the fundamental frequency. The power and value of the identified fundamental frequency was then reviewed against threshold ranges to determine whether it was in an appropriate range for running data. The filtered acceleration magnitude data was filtered again to identify the exact time location of the heel strikes. Cadence was then found as the average time between heel strikes. The magnitude of the resultant acceleration data at the identified heel strike locations were then extracted.

The data was then put through a Madgwick filter which was used to calculate the global orientation of the sensors. Known constraints to the IMeasureU mathematical model were then applied to adjust for any drift in orientation that may have been produced from the Magdwick filter process. Rotation matrices were then applied to isolate the accelerations in the forward direction. A Butterworth filter with a cut off frequency of 23Hz was then applied to the data. The data was adjusted for gravity. The algorithm then allowed us to crop the trial again twice over to ensure that we were as close as possible to the start and end of the trial (Figure 4.11 and 4.12).

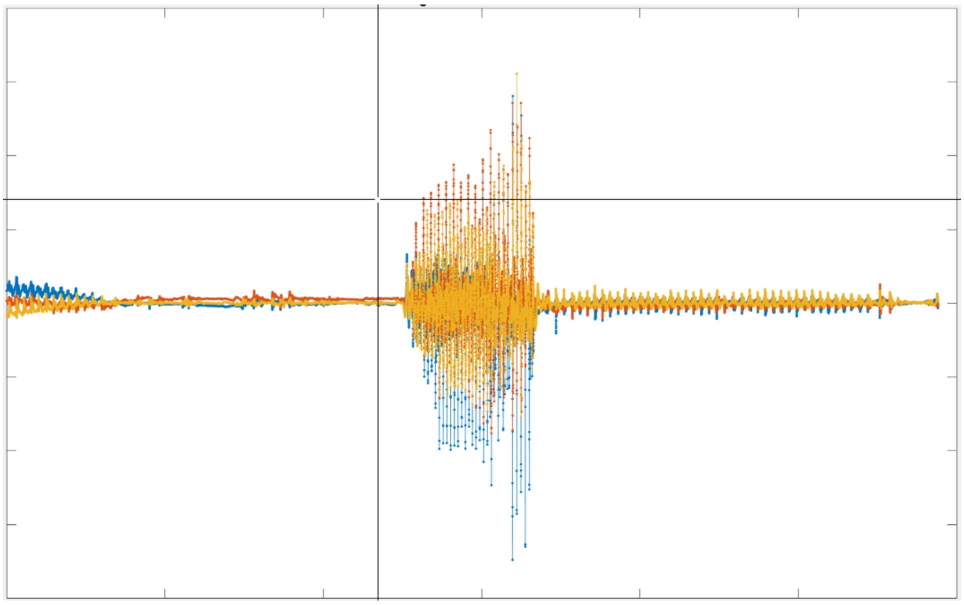

Figure 4.11 First attempt at cropping data to get closer to start and end of trial being analysed in MATLAB

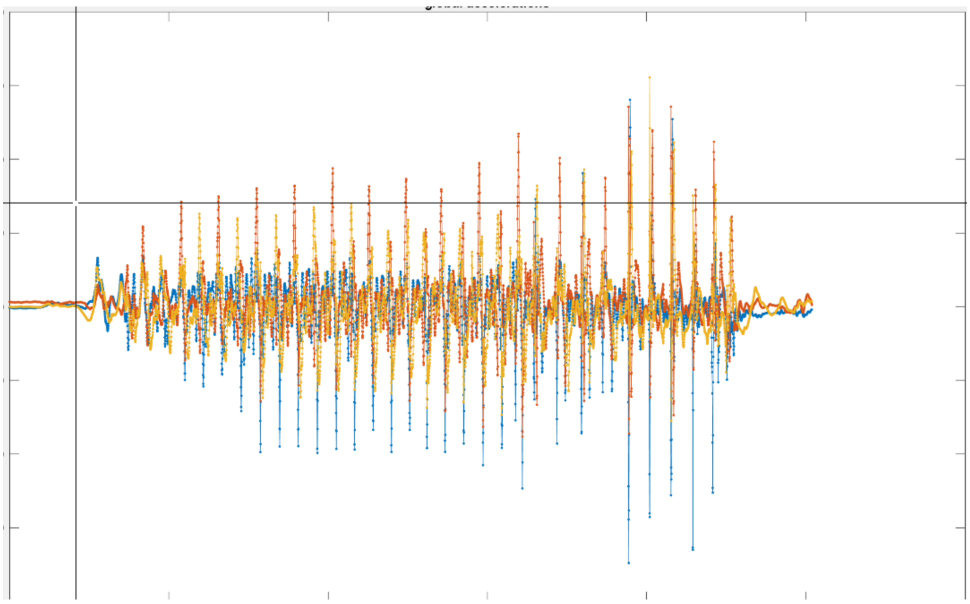

Figure 4.12 Second attempt at cropping data to get closer to start and end of trial being analysed in MATLAB

Integration was then used to calculate velocity. The drift was then removed from the original velocity using a simple linear removal (padded with zeros, and then a low pass filter was applied). Essentially, this was just removing very low frequency drift. Further integration was then used to calculate displacement. In this step, the inflection points (point at which change in direction occurs in the curve) of displacement were found. Inflection points in the displacement were at the same places where maxima and minima occurred in the velocity trace (i.e. If the gradient of the velocity was zero, the displacement value was zero). Toe offs were equal to the maximum vertical velocity from COM. The location of the heel strikes relative to the COM sensor were found. The toe offs and heel strikes were aggregated into one matrix. The distance of the run was then entered into the algorithm where horizontal velocity, contact and flight times were calculated. The final time sync data was plotted (Figure 4.13).

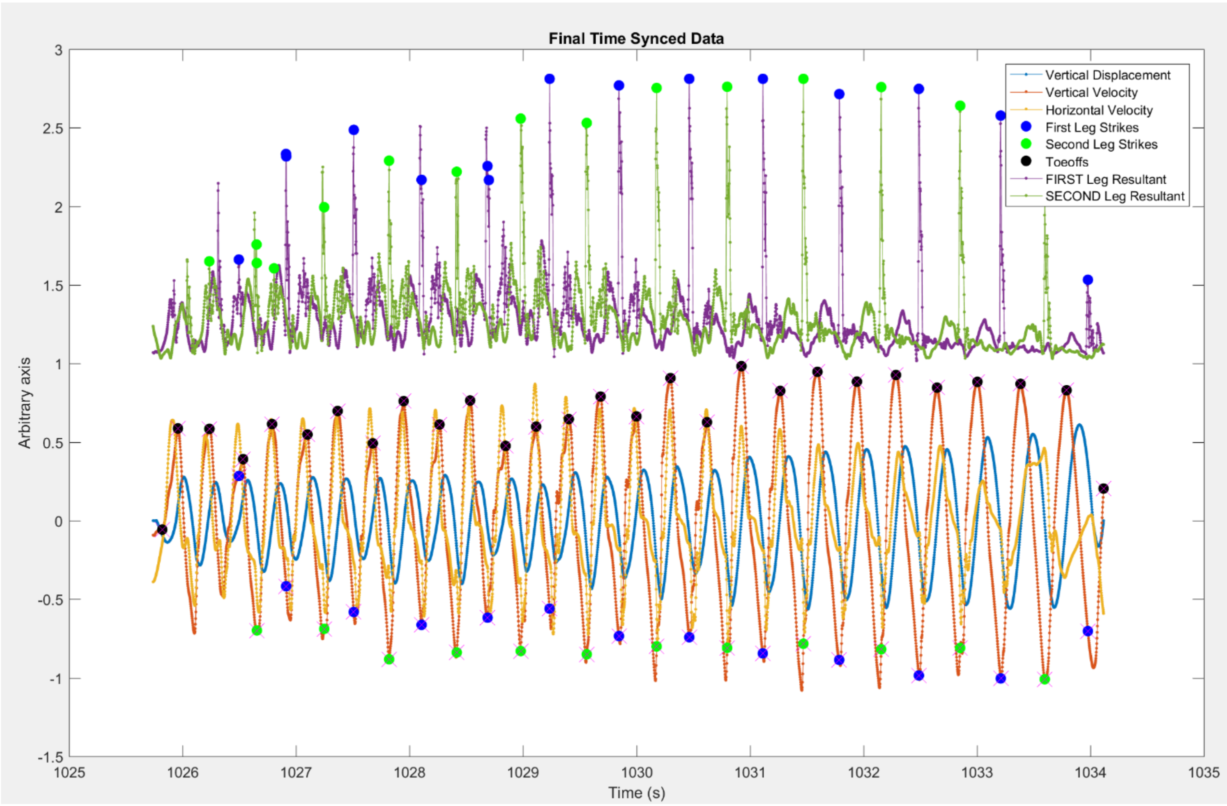

Figure 4.13 Final plot of time synced data for one trial in MATLAB

The footage of the high-speed video was then viewed to match the steps with the final sync trace and identify where the force plate step occurred. The data was then averaged using three to five contacts (dependent on dropout due to issues with the Bluetooth signal) surrounding the force plate contact. Three contacts were averaged for six trials, four contacts were averaged for twenty-two trials and five contacts were averaged for seventy-eight trials. To account for potential asymmetry the same side in which the force plate was contacted was the only side averaged (Maloney et al., 2016). For example, if the force plate contact was a left contact, the data for the two left contacts prior to the force plate step and the two left contacts following were averaged. The data was averaged as repeat contacts represent common stiffness protocol (Hobara, Inoue, Muraoka, et al., 2010; Hobara et al., 2011; Hobara et al., 2008; Hobara, Kimura, et al., 2010; Hobara et al., 2009; Pickering Rodriguez et al., 2017; Pruyn et al., 2012, 2013; Watsford et al., 2010). These contacts occurred when the participant had reached maximum velocity.

All relevant parameters were then substituted into three pre-existing stiffness formulas (Table 4.3). The leg/vertical stiffness scores from these formulas were then compared to the gold standard stiffness scores from the 3D motion analysis system.

Table 4.3 *Pre-existing formulas used to calculate IMeasureU stiffness measures*

Type of Stiffness	Formula
1. Vertical Stiffness (Dalleau et al., 2004)	$K_{vert} = \dfrac{M \times \pi(T_f + T_c)}{Tc^2\left(\frac{T_f + T_c}{\pi} + \frac{T_c}{4}\right)}$
2. Leg Stiffness (McMahon & Cheng, 1990)	$K_{leg} = \dfrac{F_{max}}{\Delta L}$ Change in leg length: $\Delta L = \Delta y + L_0(1 - Cos\theta)$ $\theta = sin^{-1}(\frac{ut_c}{2L_0})$
3. Leg Stiffness (Morin et al., 2005)	1. Estimated peak vertical force: $F_{max} = mg\frac{\pi}{2}(\frac{t_f}{t_c} + 1)$ 2. Vertical displacement of COM: $\Delta y_c = \frac{F_{max}t_{c^2}}{m\pi^2} + g\frac{{t_c}^2}{8}$ 3. Change in leg length: $\Delta L = L - \sqrt{L^2 - (\frac{vt_c}{2})^2} + \Delta y_c$ 4. Leg stiffness: $K_{leg} = \frac{F_{max}}{\Delta L}$

Note. 1. K_{vert} vertical stiffness, M body mass, T_f flight time, T_c contact time

2. K_{leg} leg stiffness, F_{max} maximum force, ΔL change in leg length, Δy maximum COM displacement, L_0 standing leg length, u horizontal velocity, t_c contact time
3. F_{max} maximum force, m body mass, g gravity, t_f flight time, t_c contact time, Δy_c maximum

4.7 STATISTICAL ANALYSIS

Prior to statistical analysis, outliers were removed using Box and Whisker plots. Outliers were defined as scores greater than 1.5x the interquartile range (Ghasemi & Zahediasl, 2012) and were removed based on the results of each calculation. Stiffness scores for each calculation were then placed into custom validity and reliability spreadsheets (Hopkins, 2015). All stiffness scores were log transformed prior to analysis.

Coefficient of variation (CV), Pearson product-moment correlations (r) and percentage differences (Bias) were used to determine the validity of the IMeasureU stiffness scores compared with the gold standard stiffness measures from the motion analysis system. Measures were deemed valid if either of the stiffness calculations displayed a correlation of greater than 0.90 and a CV of less than 10% (Atkinson & Nevill, 1998). Bland-Altman plots were also generated to assess potential bias or systematic errors with the stiffness measures differences across the range of observed scores.

To account for potential asymmetry, reliability measures for the IMeasureU data were divided into left and right contacts. ICC and CV were used to determine reliability. Measures were deemed reliable if either of the stiffness calculations or individual variables displayed an ICC of greater than 0.90 and a CV of less than 10% (Atkinson & Nevill, 1998).

CHAPTER 5: VALIDITY AND RELIABILITY OF INERTIAL MEASUREMENT UNITS IN OBTAINING A LOWER LIMB STIFFNESS MEASURE

Chapters 2 & 3 have established that lower limb stiffness is an effective mechanism in improving performance and injury prevention. However, these chapters highlighted that studies examining such links have traditionally utilised vertical hopping and jump tasks for stiffness assessment in athletic populations. These chapters also identified that stiffness assessment has traditionally involved the use of a force plate. In particular, Chapter 3, acknowledged that the longitudinal assessment of lower limb stiffness in athletic populations has so far been isolated to laboratory-based settings, which is turn has limited the regularity by which stiffness can be assessed given limitations surrounding cost, transportation and laboratory access. Therefore, not only is there a need to establish a more functional or sports- specific lower limb stiffness measure in high-level track and field athletes but also a measure that can be implemented in the daily training environment. With the future of monitoring an athlete's response to load more than likely being dominated by the emergence of new technologies, it is thought that information provided to coaches in near real time may enhance an athlete's performance and minimise their risk of injury. This Chapter addresses the validation process of using inertial measurement units to gain a measure of lower limb stiffness during running in high-level track and field athletes. It is anticipated that this study is the first step in developing a valid field-based option for sport practitioners, which will ultimately make the monitoring of athletes in their daily training environment more accessible.

5.1 ABSTRACT

The measurement of lower limb stiffness has traditionally been limited to laboratory-based settings using simple vertical hopping and jump tasks on a force plate. Current developments in inertial measurement units may allow the quantification of lower limb stiffness outside the traditional laboratory-based setting during sport specific tasks. Therefore, the primary aim of this study was to examine the validity and reliability of inertial measurement units in obtaining a lower limb stiffness running measure, comparing measures from three existing stiffness calculations with the accepted stiffness measures from a 3D motion analysis system and force plate. Nineteen high-level track and field athletes (six sprinters, seven middle distance runners and six hurdlers; Age: 22.05 ± 3.39yrs, Height: 1.71 ± 0.09m, Weight: 62.56 ± 11.66kg) performed six running gait trials at a pace reflective of their respective event during competition. Poor validity was found between the gold standard stiffness measures and the measures derived from the IMeasureU sensors. In addition, the results demonstrated that the data output from the IMeasureU sensors were not reliable when substituted into the existing measures of stiffness. Further investigation and refinement of the algorithm is required to establish its usefulness in obtaining a lower limb stiffness measure outside of the traditional laboratory-based setting.

5.2 INTRODUCTION

Lower limb stiffness quantifies the relationship between the amount of leg flexion and the load to which the limbs are subjected (Butler et al., 2003; Latash & Zatsiorsky, 1993). The quantification of lower limb stiffness is of importance to sport practitioners, coaches and athletes given its association with performance maximisation and injury risk (Watsford et al., 2010). Research has established strong links between lower limb stiffness and key performance measures including stride frequency and running velocity (Arampatzis et al., 1999; Farley & González, 1996; Kuitunen et al., 2002) as well as links between stiffness and injury risk, where higher levels of stiffness are associated with an elevated risk of overuse, bone related injuries and lower levels of stiffness related to soft tissue injury risk (Butler et al., 2003). However, current research into lower limb stiffness measures in athletes has been limited to laboratory-based settings. While laboratory-based settings offer a controlled environment for the collection of various performance measures in elite athletes, sport specific movements are often altered or omitted due to restrictions in laboratory size (Challis, 2001; Mungovan, Peralta, Gass, & Scanlan, 2018).

Measures of lower limb stiffness have most commonly been determined through simple vertical hopping and jump tasks on a force plate (Arampatzis, Schade, et al., 2001; Farley et al., 1998; Granata et al., 2002; Hobara, Inoue, Muraoka, et al., 2010; Hobara et al., 2011; Hobara et al., 2008; Hobara, Kimura, et al., 2010; Hobara et al., 2009; Pruyn et al., 2012, 2013; Watsford et al., 2010). Force plates are often used in conjunction with a fixed motion analysis system to obtain other mechanical parameters relevant to human motion (Brughelli & Cronin, 2008b). Although these measures are considered the gold standard for stiffness assessment, limited opportunity exists to measure lower limb stiffness on a regular basis in athletic populations during sport specific tasks, due to restricted laboratory access, cost and difficulty with transportation (Balsalobre-Fernández et al., 2015; Brughelli & Cronin, 2008b; Iosa et al., 2016). Therefore, alternative options such as inertial measurement units need to be considered.

A rapidly growing alternative in the analysis of human movement is the use of inertial measurement units (Hood et al., 2012). Recent developments in inertial measurement units (i.e. Wireless Bluetooth technology, increased sampling rates, storage and processing capacity) are thought to allow the tracking of athletes outside the traditional laboratory setting. Application of inertial measurement units in daily training environment monitoring has the potential for performance enhancement and injury prevention. Inertial measurement units are electronic devices that allow acceleration, heading and orientation to be captured and data logged in a small unit while fixed onto an object or human's limb segment (Seel et al., 2014). The ability to identify certain gait parameters from accelerometers and inertial measurement units has been known for some time (McCamley et al., 2012). However, the use and validation of such technology has so far been limited to stride parameters, contact times and vertical displacement (Gullstrand, Halvorsen, Tinmark, Eriksson, & Nilsson, 2009; Hobara, Inoue, Gomi, et al., 2010; Lee, Mellifont, et al., 2010; Lee, Sutter, et al., 2010; Morin et al., 2005). To date, there has been limited use of inertial measurement units in obtaining a lower limb stiffness measure, but early research suggests the application of inertial measurement units has the capacity to provide a surrogate force measure and estimate mechanical load during ground contact (Moresi, O'Meara, & Graham, 2013).

With the varying nature of sporting demands and the way an athlete deals with mechanical load thought to influence performance and injury risk, the primary aim of this study was to develop a valid and reliable measure of lower limb stiffness during running (a task reflective of training and competition) using inertial measurement units. It was hypothesised that measures of lower limb stiffness using inertial measurement units would be valid and reliable when compared to 3D motion analysis (gold standard).

5.3 METHODS

5.3.1 Research Design

The present study employed a cross-sectional study design investigating high-level track and field athletes to validate the use of inertial measurement units in obtaining a lower limb stiffness running measure.

5.3.2 Participants

Nineteen high-level track and field athletes (six sprinters, seven middle distance runners and six hurdlers; Age: 22.05 ± 3.39yrs, Height: 1.71 ± 0.09m, Weight: 62.56 ± 11.66kg) were recruited to participate in this study. Participants needed to be high-level, able-bodied athletes defined by Athletics Australia qualifying standards for relevant national team representation (i.e. Oceania Championships, World Junior Championships, World Youth Championships, Commonwealth Games, Olympic Games, World Championships and World University Games). Participants also needed to be free from a lower body injury and be completing full training at the time of testing. The study was approved by the Australian Catholic University Human Research Ethics Committee (Approval no. 2016-284H). Informed consent was obtained from all participants prior to participation.

5.3.3 Testing Procedure

Following a self-directed warm up, which consisted of a full body activity (i.e. jogging, cycling), dynamic stretching and event specific drills, participants were instructed to perform six running gait trials from a standing start at a pace reflective of their event during competition. There were no restrictions placed on the participants starting position. However, running speeds were monitored using timing gates placed 20m apart in the testing area to ensure consistency between trials. The distance of every run was also measured to input into the MATLAB algorithm. The distances run ranged between 40 and 51m. Along with ensuring consistent running speeds, participants needed to attain six-foot strikes on the force plate (i.e. three left foot contacts and three right foot contacts). Participants were required to wear tight bike shorts, crop top (females) and running spikes to allow adequate exposure of all relevant markers and IMeasureU sensor placement. Running spikes were worn to mimic an athlete's training and competition environment. Adequate recovery time was provided between each trial to avoid the influence of fatigue on the required stiffness measures.

All testing was performed on an indoor 55m Mondo running track in the biomechanics laboratory at the New South Wales Institute of Sport. A fourteen camera, 3D motion analysis system (VICON Vantage 5; Oxford Metrics Ltd, Oxford, United Kingdom) sampling at 250Hz was used to determine full body kinematic data during each trial. Four floor mounted force plates (Kistler 9281CA, Winterthur, Switzerland) (situated 30m along the 55m running track) sampling at a rate of 1000Hz were used to enable the necessary assessment and measurement of ground reaction forces during each trial. High speed video was captured for participant feedback as well as to determine when and where each foot strike occurred. Data was captured simultaneously using three IMeasureU Blue Thunder sensors (accelerometer: 3-axis range of ±16g; gyroscope: 3-axis full scale range of ±2000°/sec; magnetometer: 3-axis full scale range of ±1200mT) (IMeasureU, Auckland, New Zealand) (Firmware Version 1.3) and Research App for iPhone (IMeasureU Research App, Version 3.3). The IMeasureU sensors sampling at a rate of 500Hz were placed on both the left and right medial distal tibias as well as in the middle of the left and right PSIS markers for each participant (i.e. approximate COM location) (Appendix F). The IMeasureU sensor placed on the mid PSIS was secured with medical adhesive tape while a Velcro strap attached both tibia sensors. All IMeasureU sensors were placed on the participant following marker placement and were attached without discomfort or in a way that restricted movement but ensured minimal artifactual movement of the sensor. The

IMeasureU Research App was used to log data to an on-board micro SD card for each sensor as well as to enter the start and stop times for each trial. Once all the trials were completed, the .csv files for each individual sensor were exported from the Research App via email. Each IMeasureU sensor was then plugged into the computer and the .csv files were synced with the data stored onboard each unit through the Lightning Desktop App (Version 3.0.0). It should be noted that the IMeasureU sensors were not synced with Vicon due to the IMeasureU-Vicon sync bandwidth approach (i.e. the accelerometer, magnetometer and gyroscope data (9axis) can only be captured and synced at a sample frequency of 100Hz).

5.3.4 Data Processing

5.3.4.1 Motion Analysis Data

A dual low pass, fourth order Butterworth filter with a cut off frequency of 23Hz was used to smooth the kinematic data (Millett et al., 2014). Following filtering, the plug-in gait model was used to calculate force and COM displacement. Vertical stiffness was calculated as PVGRF divided by maximum vertical COM displacement and leg stiffness was calculated as the maximum vertical force divided by the change in vertical leg length (McMahon & Cheng, 1990). Leg length change accounts for COM, horizontal velocity at touchdown and contact time. Only the eccentric displacement during landing was used to calculate stiffness.

5.3.4.2 IMeasureU Data

The raw IMeasureU data for each participant were imported into a custom-made MATLAB algorithm (R2016b, The Mathworks Inc, Natick, MA), in which all three sensors were synced together. The algorithm was configured to receive and process the raw 3 axis acceleration data and calculate the resultant acceleration magnitude data at each time sample. The peak shock for each individual foot strike was determined by the magnitude of the resultant acceleration vector at the location of the heel strike in each foot strike, as this is where the maximum shock occurs in the foot strike. This was calculated by square rooting the sum of the squares for all three acceleration measures at each time point. The resultant magnitude data was then filtered using a bandpass filter to filter out high frequency noise (i.e. skin movement) and very low frequency movement. The filtered resultant acceleration data was then processed to identify the fundamental frequency. A fast Fourier Transform was used to find the fundamental frequency. The power and value of the identified fundamental frequency was then reviewed against threshold ranges to determine whether it was in an appropriate range for running data. The filtered acceleration magnitude data was filtered again to identify the exact time location of the heel strikes. Cadence was then found as the average time between heel strikes. The magnitude of the resultant acceleration data at the identified heel strike locations were then extracted. The data was then put through a Madgwick filter which was used to calculate the global orientation of the sensors. Known constraints to the IMeasureU mathematical model were then applied to adjust for any drift in orientation that may have been produced from the Magdwick filter process. Rotation matrices were then applied to isolate the accelerations in the forward direction. A Butterworth filter with a cut off frequency of 23Hz was then applied to the data. The data was adjusted for gravity. Integration was used to calculate velocity. Further integrations were used to calculate displacement. The distance of the run was entered into the algorithm where horizontal velocity, contact and flight times were calculated. The final time sync data was plotted.

The footage of the high-speed video was viewed to match the steps with the final sync trace produced from the algorithm to determine where the force plate contact occurred. The data was then averaged using three to five contacts (dependent on dropout due to issues with Bluetooth signal) surrounding the force plate contact. Three contacts were averaged for six trials, four contacts were averaged for twenty-two trials and five contacts were averaged for seventy-eight trials. To account for potential asymmetry, the same side in which the force plate was contacted was the only side averaged (Maloney et al., 2016). For example, if the force plate contact was a left contact, the data for the two left contacts prior to the force plate step and the two left contacts following were averaged. The data was averaged as repeat contacts represent common stiffness protocol (Hobara, Inoue, Muraoka, et al., 2010; Hobara et al., 2011; Hobara et al., 2008; Hobara, Kimura, et al., 2010; Hobara et al., 2009; Pickering Rodriguez et al., 2017; Pruyn et al., 2012, 2013; Watsford et al., 2010). These contacts occurred when the participant had reached maximum velocity.

All relevant parameters were then substituted into three pre-existing stiffness formulas (Table 5.1). The vertical and leg stiffness scores from these formulas were then compared to the gold standard stiffness scores from the motion analysis system.

Table 5.1 *Pre-existing formulas used to calculate IMeasureU stiffness measures*

Type of Stiffness	Formula
1. Vertical Stiffness (Dalleau et al., 2004)	$K_{vert} = \dfrac{M \times \pi(T_f + T_c)}{Tc^2\left(\dfrac{T_f + T_c}{\pi} + \dfrac{T_c}{4}\right)}$
2. Leg Stiffness (McMahon & Cheng, 1990)	$K_{leg} = \dfrac{F_{max}}{\Delta L}$ Change in leg length: $\Delta L = \Delta y + L_0(1 - Cos\theta)$ $\theta = sin^{-1}(\dfrac{ut_c}{2L_0})$
3. Leg Stiffness (Morin et al., 2005)	1. Estimated peak vertical force: $F_{max} = mg\dfrac{\pi}{2}(\dfrac{t_f}{t_c} + 1)$ 2. Vertical displacement of COM: $\Delta y_c = \dfrac{F_{max}t_{c^2}}{m\pi^2} + g\dfrac{{t_c}^2}{8}$ 3. Change in leg length: $\Delta L = L - \sqrt{L^2 - (\dfrac{vt_c}{2})^2 + \Delta y_c}$ 4. Leg stiffness: $K_{leg} = \dfrac{F_{max}}{\Delta L}$

Note. 1. K_{vert} vertical stiffness, M body mass, T_f flight time, T_c contact time

2. K_{leg} leg stiffness, F_{max} maximum force, ΔL change in leg length, Δy maximum COM displacement, L_0 standing leg length, u horizontal velocity, t_c contact time
3. F_{max} maximum force, m body mass, g gravity, t_f flight time, t_c contact time, Δy_c maximum vertical COM displacement, ΔL change in leg length, L initial leg length

5.3.5 Statistical Analysis

Prior to statistical analysis, outliers were removed using Box and Whisker plots. Outliers were defined as scores greater than 1.5x the interquartile range (Ghasemi & Zahediasl, 2012) and were removed based on the results of each calculation. Stiffness scores for each calculation were then placed into custom validity and reliability spreadsheets (Hopkins, 2015). All stiffness scores were log transformed prior to analysis.

CV, Pearson product-moment correlations (r) and percentage differences (Bias) were used to determine the validity of the IMeasureU stiffness scores compared with the gold standard stiffness measures from the motion analysis system. Measures were deemed valid if either of the stiffness calculations displayed a correlation of greater than 0.90 and a CV of less than 10% (Atkinson & Nevill, 1998). Bland-Altman plots were also generated to assess potential bias or systematic errors with the stiffness measures differences across the range of observed scores.

To account for potential asymmetry, reliability measures for the IMeasureU data were divided into left and right contacts. ICC and CV were used to determine reliability. Measures were deemed reliable if either of the stiffness calculations or individual variables displayed an ICC of greater than 0.90 and a CV of less than 10% (Atkinson & Nevill, 1998).

5.4 RESULTS

Overall, data from 114 sprints were collected. However, due to unforeseen technological issues (i.e. drop out of contacts from IMeasureU sensors), only data from 106 sprints could be analysed. These 106 sprints were made up of 54 right force plate contacts and 52 left force plate contacts before outliers were removed.

5.4.1 Validity

Poor validity was evident across all measures of vertical and leg stiffness when compared to the gold standard stiffness measures from the motion analysis system (Table 5.2). All of the measures displayed extreme bias and a large typical error of measurement (CV%: 30.9 – 32.7). Both leg stiffness measures reported no relationship between the two measures, while a moderate relationship was found for the vertical stiffness measure.

Table 5.2 *Validity results for 3D motion analysis Vs. IMeasureU stiffness measures*

	Vertical Stiffness (Dalleau et al., 2004)	**Leg Stiffness** (McMahon & Cheng, 1990)	**Leg Stiffness** (Morin et al., 2005)
Mean Bias %	81.8	88380269.2	86.8
CV%	31.5	32.7	30.9
r	0.52	-0.02	0.09

The Bland Altman plots from the raw data suggest that Dalleau et al. (2004) (Figure 5.1) and Morin et al. (2005) (Figure 5.2) stiffness measures report similar values. However, both measures still appear to have a proportional bias towards the higher stiffness scores. The McMahon and Cheng (1990) (Figure 5.3) measures also appear to have a moderate proportional bias towards the higher residuals with higher scores. The limits of agreement were calculated as 95% for each calculation.

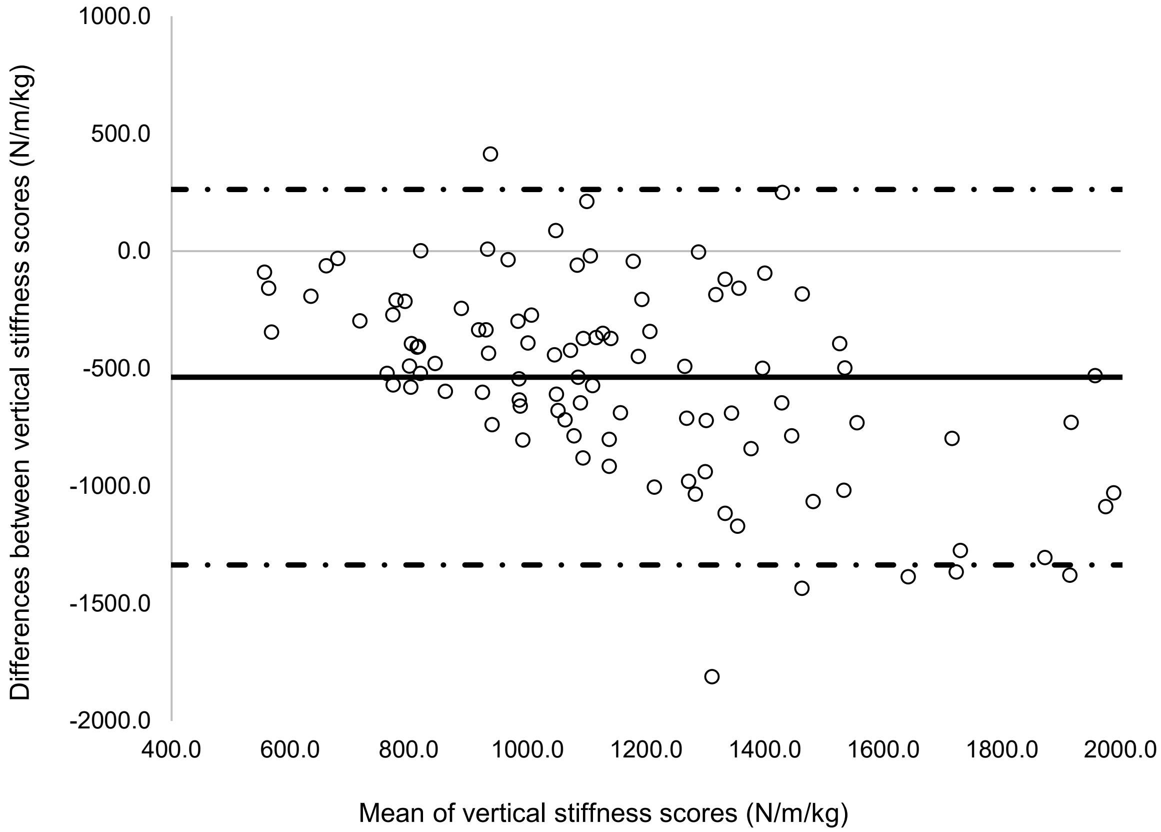

Figure 5.1 3D motion analysis raw vertical stiffness scores Vs. IMeasureU raw vertical stiffness scores (Dalleau et al., 2004)

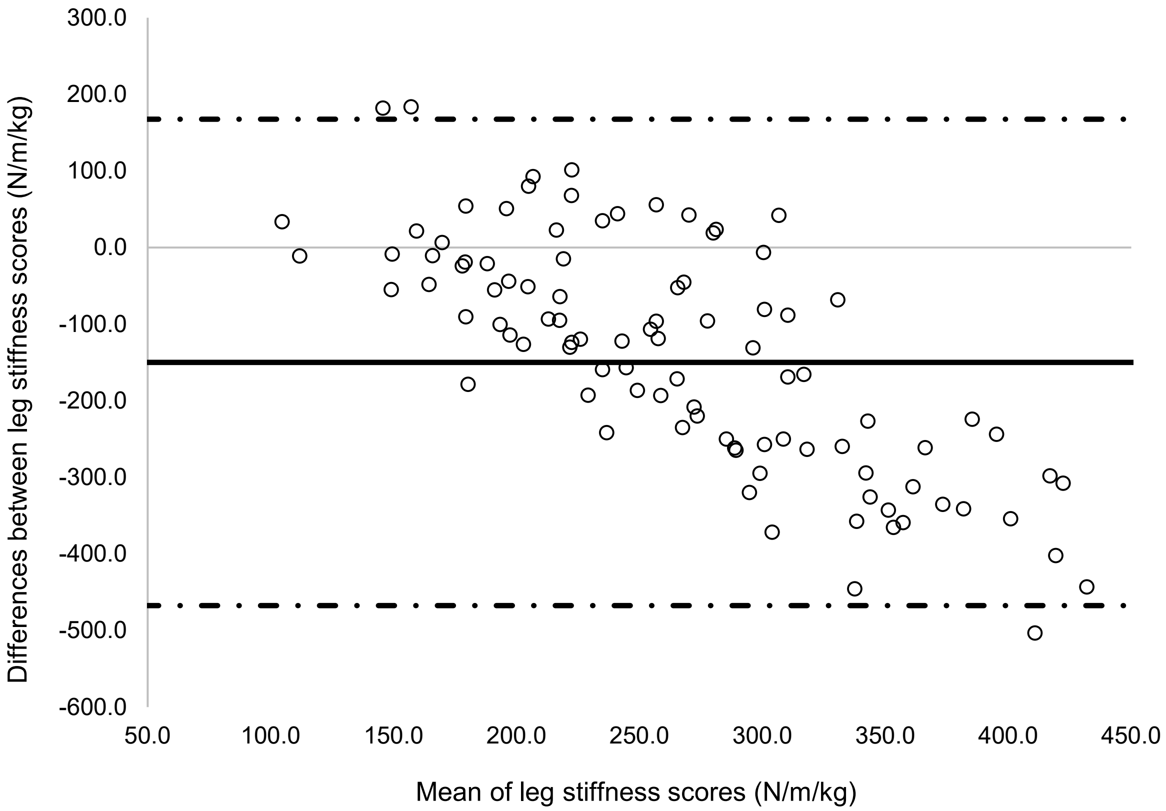

Figure 5.2 3D motion analysis raw leg stiffness scores Vs. IMeasureU raw leg stiffness scores (Morin et al., 2005)

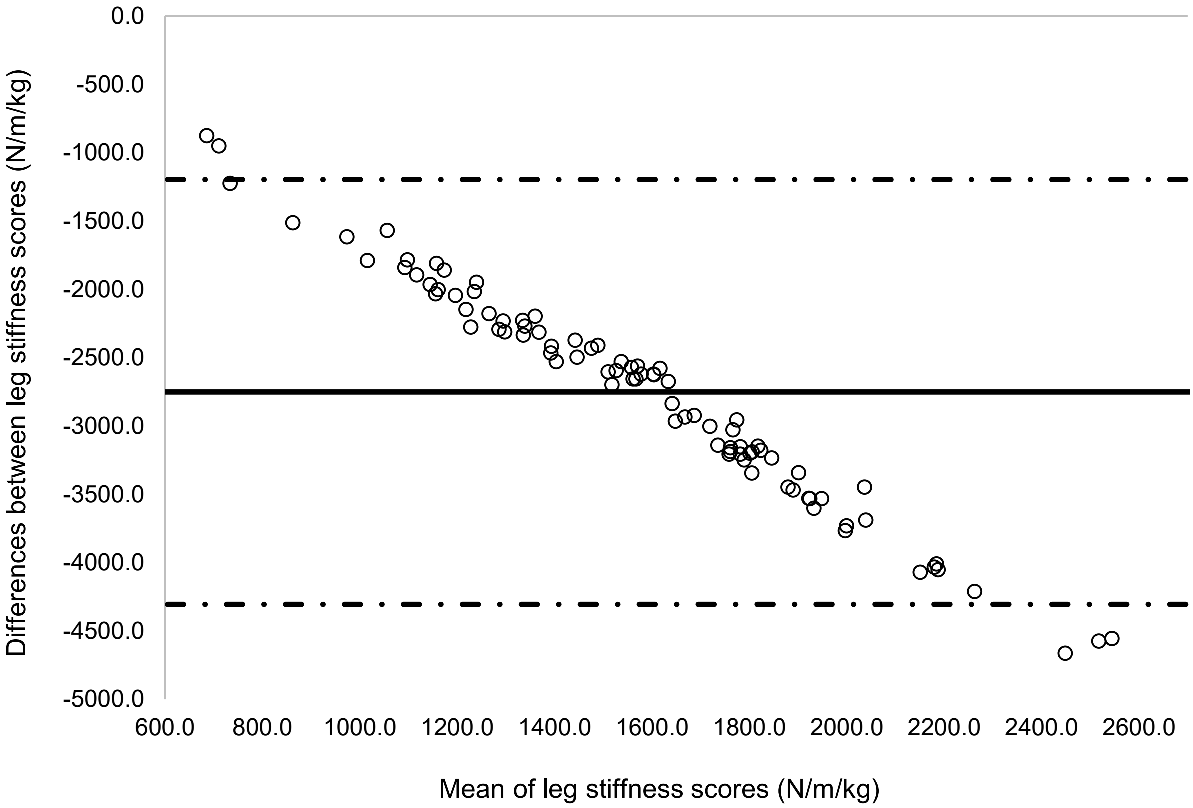

Figure 5.3 3D motion analysis raw leg stiffness scores Vs. IMeasureU raw leg stiffness scores (McMahon & Cheng, 1990)

The validity of each individual variable derived from the 3D motion analysis and IMeasureU sensors are shown in Table 5.3. Horizontal velocity was found to be the only valid variable (CV%: 5.5 - 5.6, ICC: 0.92 - 0.94). Contact time, velocity, force and COM displacement variables were underestimated when compared to the 3D motion analysis system. However, flight time was overestimated.

Table 5.3 *Validity results of individual variables for 3D motion analysis Vs. IMeasureU data*

Variable	Validity Measure	Vertical Stiffness (Dalleau et al., 2004)	Leg Stiffness (McMahon & Cheng, 1990)	Leg Stiffness (Morin et al., 2005)
Contact Time	Mean Bias %	-7.8	-4.5	-7.5
	CV%	18.7	16.7	18.8
	r	0.36	0.41	0.35
Flight Time	Mean Bias %	6.2		5.9
	CV%	13.6		13.7
	r	0.45		0.45
Horizontal Velocity	Mean Bias %		-20.6	-20.9
	CV%		5.6	5.5
	r		0.92	0.94
Force	Mean Bias %		-59.0	
	CV%		23.0	
	r		0.55	
COM Displacement	Mean Bias %		-20.8	
	CV%		25.4	
	r		0.04	

5.4.2 Reliability

Reliability for all the stiffness measures derived from the IMeasureU sensors were below the ICC cut-off of 0.90 and above the CV cut-off of 10% (Table 5.4).

Table 5.4 *Reliability measures for IMeasureU data left vs. right contacts*

	Force Plate Contact Side	**Vertical Stiffness** (Dalleau et al., 2004)	**Leg Stiffness** (McMahon & Cheng, 1990)	**Leg Stiffness** (Morin et al., 2005)
CV%	Left	13.5	15.3	28.0
	Right	18.6	23.0	39.7
ICC	Left	0.68	0.60	0.56
	Right	0.68	0.33	0.42

The reliability for each of the individual variables derived from the IMeasureU sensors required for each stiffness calculation are presented in Table 5.5. Horizontal velocity was found to be the only reliable variable for both limbs of the McMahon and Cheng (1990) calculation (CV%: 4.0 - 4.8, ICC: 0.90 - 0.91) and the left limb of the Morin et al. (2005) calculation (CV% 4.3, ICC: 0.90). Contact time for left limb of the Dalleau et al. (2004) calculation, velocity for the right limb of the Morin et al. (2005) calculation and force for both limbs of the McMahon and Cheng (1990) calculation were the only other variables to display a CV% below 10%. On the other hand, COM displacement was the only variable to display a CV% above 20% (36.6 - 41.2%) indicating very poor reliability.

Table 5.5 *Reliability measures for individual variables required for each IMeasureU stiffness calculation*

Variable	Force Plate Contact Side	Reliability Measure	Vertical Stiffness (Dalleau et al., 2004)	Leg Stiffness (McMahon & Cheng, 1990)	Leg Stiffness (Morin et al., 2005)
Contact Time	Left	CV%	9.3	12.5	10.0
		ICC	0.61	0.48	0.64
	Right	CV%	13.7	15.5	13.1
		ICC	0.59	0.56	0.41
Flight Time	Left	CV%	10.8		10.4
		ICC	0.74		0.75
	Right	CV%	14.5		14.5
		ICC	0.78		0.59
Horizontal Velocity	Left	CV%		4.0	4.3
		ICC		0.90	0.90
	Right	CV%		4.8	6.0
		ICC		0.91	0.83
Force	Left	CV%		5.3	
		ICC		0.44	
	Right	CV%		5.3	
		ICC		0.71	
COM Displacement	Left	CV%		42.1	
		ICC		0.54	
	Right	CV%		36.6	
		ICC		0.53	

5.5 DISCUSSION

This study assessed the validity and reliability of IMeasureU sensors in obtaining a measure of lower limb stiffness during running. We compared measures from three existing stiffness calculations with the gold standard (stiffness measures) from a 3D motion analysis system. Our findings show that poor validity exists between the gold standard stiffness measures and the measures derived from the IMeasureU sensors. In addition, our results demonstrated that the data output from the IMeasureU sensors were not reliable when substituted into the existing measures of stiffness, warranting the need for further investigation.

Based on previous literature, data derived from the IMeasureU sensors were substituted into three existing stiffness formulas that were split into measures of vertical (Dalleau et al., 2004) and leg stiffness (McMahon & Cheng, 1990; Morin et al., 2005). Of these measures, vertical stiffness proved to be the most favourable measure of stiffness in terms of validity and reliability. This calculation only required the input of contact and flight time variables derived from the IMeasureU sensors (Dalleau et al., 2004). However, our findings demonstrated that both of these variables were not valid which contrasts the findings of previous literature. A recent study using inertial sensors found ground contact times to be valid across a range of running velocities in comparison to a high-speed video camera (Ammann et al., 2016). Another study looking at the validation of temporal gait parameters using inertial measurement units also found valid ground contact times during walking in comparison to a force plate (Patterson et al., 2016). However, these studies attributed the high accuracy of ground contact times to the positioning of the sensors on an individual's feet/shoes (Ammann et al., 2016; Patterson et al., 2016). It has been suggested that the closer the sensors are located to the feet, the more accurate the detection of ground contact time due to less attenuation of ground reaction forces (Patterson & Caulfield, 2010). Therefore, improvements in the validity and reliability of these variables may lie in the repositioning of the sensors from the tibia to the distal aspect of both feet/shoes.

While improvements around the positioning of the sensors may hold true, previous research has outlined measurement errors in the detection of temporal parameters (i.e. ground contact time) using inertial measurement units. A recent study, attempting to estimate temporal parameters during sprint running using a trunk mounted inertial sensor, was unable to determine any consistently identifiable feature that correlated with ground contact and flight time (Bergamini et al., 2012). However, earlier research was able to successfully identify ground contact and toe off using a trunk mounted accelerometer and video pictures in endurance runners (Auvinet, Gloria, Renault, & Barrey, 2002). Therefore, it has been suggested that the varying nature of the task (i.e. sprinting vs. distance running) may lead to differences in damping, shock and vibration which in turn may affect the measurement accuracy of inertial measurement units (Bergamini et al., 2012).

When comparing the resultant acceleration traces of the sprinters and middle-distance runners included in our research, there were no clear differences in how each contact point was being detected. However, looking at the acceleration traces in more detail, it was evident that there was a double peak in the trace for the middle-distance runners. It is believed that this double peak is a representation of a mid/rearfoot strike, which concurs with the earlier findings of Cavanagh and Lafortune (1980) who demonstrated that the vertical ground reaction force pattern in rearfoot strikers displayed a double peak. Further examination from the high-speed video, also showed that the middle distance runners landed with their feet in an abducted position (i.e. toes facing outward) which is consistent with the research of Cavanagh et al. (1985). Therefore, we believe that the first peak in the acceleration trace should indicate ground contact. Given the middle-distance runners have more medial-lateral movement, the second and highest peak should occur once they have already contacted the ground. Currently, the algorithm determines ground contact for the middle-distance runners as the second and highest peak which may provide an explanation as to why contact times are underestimated, and flight times are overestimated. However, due to time constraints associated with the completion of a Master's degree it was not within the scope to adapt the algorithm to account for individualised variations in contact. Therefore, it is acknowledged that a limitation of this study is the ability of the algorithm to correctly identify the initial peak as ground contact for rearfoot strikers. Further research is needed to determine the precision and accuracy of not only the IMeasureU sensors but other inertial measurement units and their ability to consistently identify ground contact dependent on the task and differences associated with an individual's gait pattern.

Despite, contact and flight time being important variables in the calculation of vertical stiffness, additional variables are needed for the calculation of leg stiffness. The McMahon and Cheng (1990) leg stiffness calculation is considered the accepted stiffness measure and required the input of the most variables from the IMeasureU data. Horizontal velocity was found to be the only valid and reliable variable. This finding can be attributed to the distance of each run being measured and inputted into the algorithm. COM displacement showed the poorest validity and reliability of all the variables. However, we acknowledge this finding to be the result of the limitation surrounding accurate detection of ground contact time. Our surrogate force measure (peak resultant acceleration from tibia sensors) showed a high CV (23%) for validity. In terms of reliability it displayed a CV below 10%, however the ICC fell below 0.90 for both limbs. The peak resultant acceleration was used instead of the peak acceleration due to the saturation of the x axis, which aligned with the direction of the runner. We theorise that this may be the result of braking forces.

Finally, the Morin et al. (2005) leg stiffness measure displayed poor validity and reliability. In terms of reliability, the left leg was found to be slightly more reliable than the right leg. This can be attributed to the reliability of contact time, flight time and horizontal velocity being more reliable on the left side. However, this does not provide a plausible explanation as to why it had the poorest reliability overall. A potential explanation for this may be related to the fact that previous research has found this sine-wave method to underestimate leg stiffness during hopping and running tasks when compared to other published stiffness models (Coleman, Cannavan, Horne, & Blazevich, 2012; Hobara, Inoue, Kobayashi, & Ogata, 2014).

Given the novel approach of this study there are a number of limiting factors that need to be considered. Firstly, the positioning of ground mounted force plates in the laboratory limited the amount of contacts captured by the 3D motion analysis system to one per trial. Secondly, while this was a result of factors beyond our control, technological difficulties (i.e. dropout of contacts due to issues with the Bluetooth signal) meant that eight trials couldn't be analysed. In addition, the drop out of contacts did not allow all five contacts surrounding the force plate to be averaged for twenty-eight trials. However, it should be noted that the IMeasureU sensors and Research App are constantly undergoing significant improvements in their software development. Finally, the algorithm and the way ground contact (i.e. heel strikes) are currently derived need to be reconsidered. Improvements are needed to better detect events based on an individual's gait pattern and task being performed.

5.6 CONCLUSION

The results from the current study found that the stiffness measures derived from the IMeasureU sensors were not valid when compared with the gold standard stiffness measures from the 3D motion analysis system. In addition, our results demonstrated that the data output from the IMeasureU sensors were not reliable when substituted into the existing measures of vertical/leg stiffness. Therefore, the use of IMeasureU sensors to obtain a field-based measure of lower limb stiffness cannot not be recommended. Further investigation and refinement of the algorithm is required to establish its usefulness in obtaining a lower limb stiffness measure outside of the traditional laboratory-based setting.

CHAPTER 6: DISCUSSION

6.1 OVERVIEW

The aim of this thesis was to develop a valid and reliable novel sensor-based measure to quantify lower limb stiffness in high-level track and field athletes during running. This section will summarise the key findings and implications of each study as well as provide recommendations for future research.

6.2 STUDY 1: LONGITUDINAL ASSESSMENT OF LOWER LIMB STIFFNESS IN ADULT ATHLETIC POPULATIONS: A SYSTEMATIC REVIEW

6.2.1 Key Findings

Chapter 3 systematically reviewed the existing literature to offer valuable insight into investigations already conducted and provide future direction into the longitudinal monitoring of lower limb stiffness in athletic populations. Based on the studies included for review (Nagahara & Zushi, 2017; Pickering Rodriguez et al., 2017; Pruyn et al., 2012, 2013; Serpell, Scarvell, et al., 2014; Watsford et al., 2010), long-term stiffness measures have so far been isolated to laboratory-based settings, performed on an irregular basis and predominately assessed through simple vertical hopping and jump tasks in the specific sporting population of AFL players. Currently, stiffness assessment hasn't been reported to occur as frequently as daily/weekly, which may offer an explanation as to why limited long-term stiffness changes have been found. This is a possible limitation of the existing literature as a change may occur and be subsequently missed as a result of infrequent measures. Therefore, the relevance of existing knowledge and its practical application for coaches and athletes is limited due to the lack of information based on the assessment of sport specific tasks and evaluation of diverse athletic populations on a regular basis.

6.2.2 Research Implications

This review highlighted the lack of research focusing on long-term stiffness measures in adult athletic populations during sport specific tasks. Due to the lack of longitudinal tracking and regular stiffness assessment using sport specific measures, prospective links between stiffness, performance and injury remain unclear. However, it is thought that the use of sport specific measures may provide a clearer insight into such links. Therefore, the need for a field-based measure of lower limb stiffness was identified in order for stiffness to be assessed at more regular intervals to better understand the prospective links between lower limb stiffness, performance and injury during sport specific tasks.

6.3 STUDY 2: VALIDITY AND RELIABILITY OF INERTIAL MEASUREMENT UNITS IN OBTAINING A LOWER LIMB STIFFNESS MEASURE

6.3.1 Key Findings

Chapter 5 highlighted the validity and reliability of IMeasureU sensors in obtaining a potential field-based measure of lower limb stiffness during running. Results demonstrated that the stiffness measures derived from the IMeasureU sensors were not valid when compared to the gold standard stiffness measures from the 3D motion analysis system. In addition, reliability for all stiffness measures derived from the IMeasureU sensors fell below the ICC cut-off of 0.90 and were above the CV cut-off of 10%. However, of all the measures, vertical stiffness (Dalleau et al., 2004) proved to be the most favourable stiffness measure. In terms of the individual variables that make up each stiffness calculation, horizontal velocity was found to be the only valid and reliable variable.

6.3.2 Research Implications

The use of inertial measurement units in the analysis of human movement are a rapidly growing alternative given their versatility and portability. However, measures derived from inertial measurement units must be both valid and reliable before they can be considered for use in the analysis of human movement. Currently, the IMeasureU sensors and associated algorithm are not able to determine a valid or reliable measure of vertical/leg stiffness. Further research surrounding how event (sprinters vs middle distance) and individual differences influence the detection of various parameters are needed.

6.4 RECOMMENDATIONS AND LIMITATIONS FOR CURRENT AND FUTURE RESEARCH

Based on the current findings (presented in Chapter 5), the need for further investigation is warranted. Therefore, this section will outline recommendations to improve the validity and reliability of IMeasureU sensors in obtaining a lower limb stiffness running measure. It will also acknowledge the limitations that surround the current findings. It should be noted that while initial refinement of the algorithm occurred (Appendix F), due to time constraints associated with the completion of a Master's degree it was not within the scope to adapt the algorithm to account for individualised variations in contact. However, we have outlined where improvements to the algorithm need to be made for future research to assist with the possibility of developing a valid and reliable lower limb stiffness monitoring tool.

6.4.1 Recommendation 1: Positioning of the IMeasureU sensors

Previous research looking at the validation of ground contact time from inertial measurement units has demonstrated that the position of the unit may influence the accurate detection of ground contact time (Ammann et al., 2016; Patterson et al., 2016). A recent study looking at the validation of three inertial measurement unit locations (i.e. trunk, shank and feet) in comparison to a force plate, found that the inertial measurement units placed on the feet were found to be the most accurate in comparison to the force plate due to less attenuation of ground reaction forces (Patterson et al., 2016). This finding was also confirmed by Ammann et al. (2016) who believed that the high measurement accuracy of ground contact time in comparison to high speed video and Optojump was the result of having the sensors fixed to the laces of both shoes. Therefore, one possible suggestion for improvement in the detection of ground contact time may be changing the position of the tibia sensors to the distal part of both feet/shoes.

While this may hold true for the detection of ground contact time, the detection of toe-off has not been frequently investigated in current literature. However, a recent study by Mo and Chow (2018) found that the placement of the inertial measurement unit on the vertical shank displayed the most accurate detection of toe-off when compared to a force plate during overground running. It is believed that the sudden forward and upward movement of the shank produced by the hip and knee may contribute to its accuracy (Mo & Chow, 2018). The same study examined the accuracy of toe-off from a lumbar positioned inertial measurement unit, however, it produced the largest mean difference (Mo & Chow, 2018). Despite these findings, this study identified several limitations; most notably that an individual's foot strike pattern wasn't determined. Further research surrounding the positioning of inertial measurement units during overground running is needed in order to determine the accurate detection of ground contact and toe-off based on an individual's foot strike pattern.

6.4.2 Recommendation 2: Improve detection of ground contact

Previous research has suggested that running speed, distance, training level, footwear and mechanical properties may influence an individual's foot strike pattern

(Hatala, Dingwall, Wunderlich, & Richmond, 2013; Murphy, Curry, & Matzkin, 2013). More specifically, endurance running has most commonly been associated with a rear/mid-foot strike, whereas sprinters are most commonly known for forefoot striking (Hamill, 2012). However, it remains unclear in current research whether an individual's gait pattern influences the accuracy and precision of inertial measurement units in their ability to correctly identify the point of ground contact.

Our research involved the inclusion of both sprinters and middle-distance runners. Due to the shifting orientation of the sensors, especially in the distal tibia sensor, the resultant acceleration trace was used as it was believed to best detect the peak acceleration due to contact with the ground. Figure 6.1 displays an example of a resultant acceleration trace of a sprinter and Figure 6.2 is an example of an acceleration trace for a middle-distance runner included in this research.

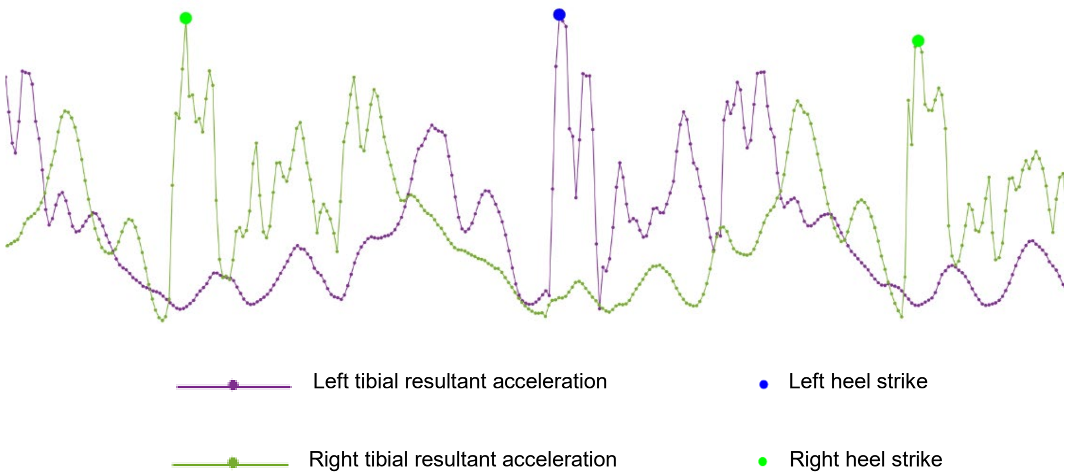

Figure 6.1 Resultant acceleration trace of a sprinter for three heel strikes

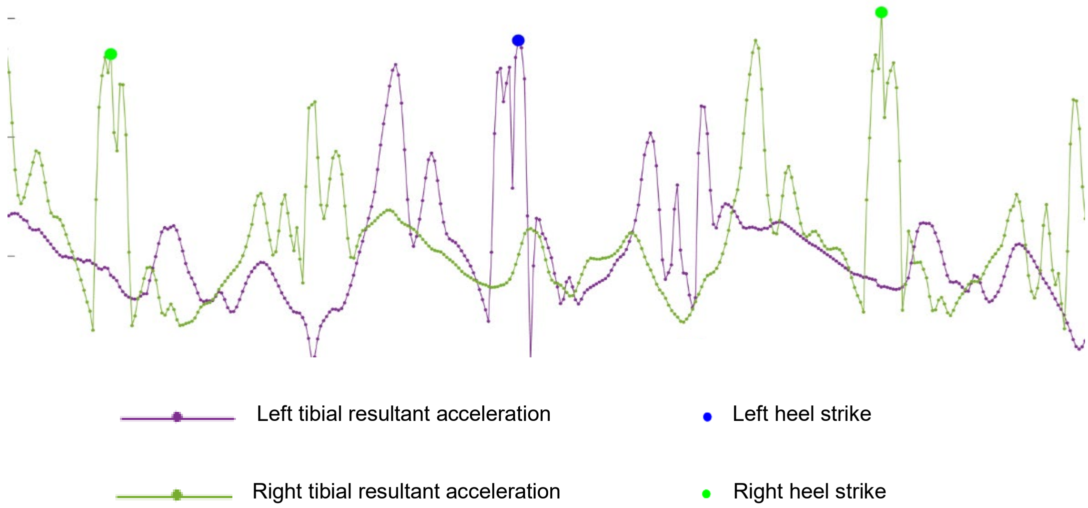

Figure 6.2 Resultant acceleration trace of a middle-distance runner for three heel strikes

When looking at the two traces there is no obvious differences in how ground contact was being detected (indicated by blue and green dots) between the sprinter and middle-distance runner. However, when examining the traces in more detail (Figure 6.3 and 6.4) there was a clear double peak in the acceleration trace of the middle-distance runner (Figure 6.4).

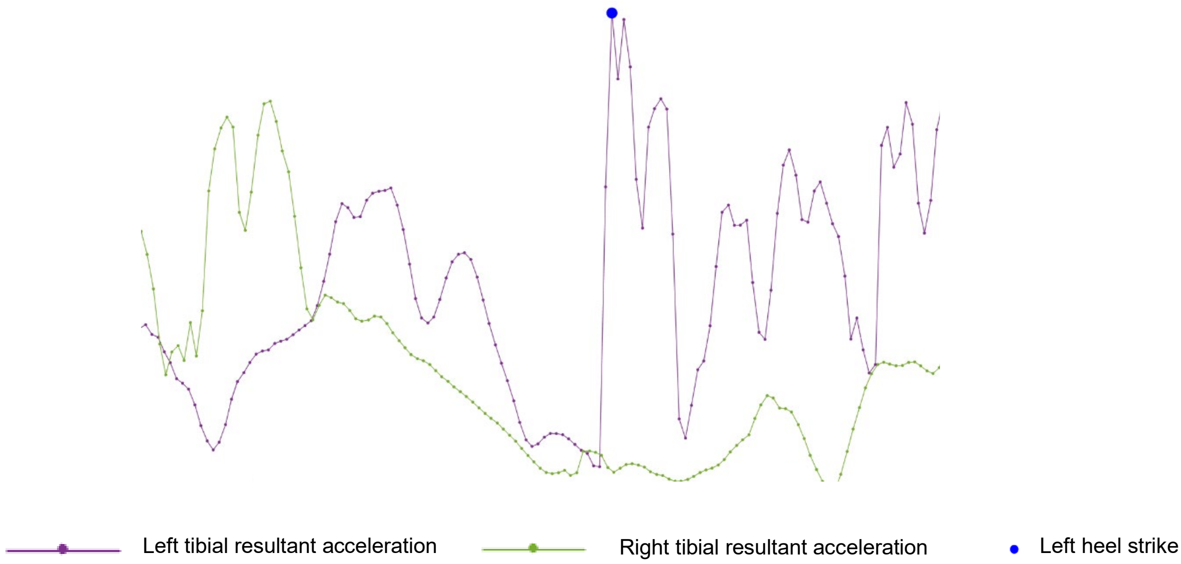

Figure 6.3 Resultant acceleration of a sprinter for one heel strike

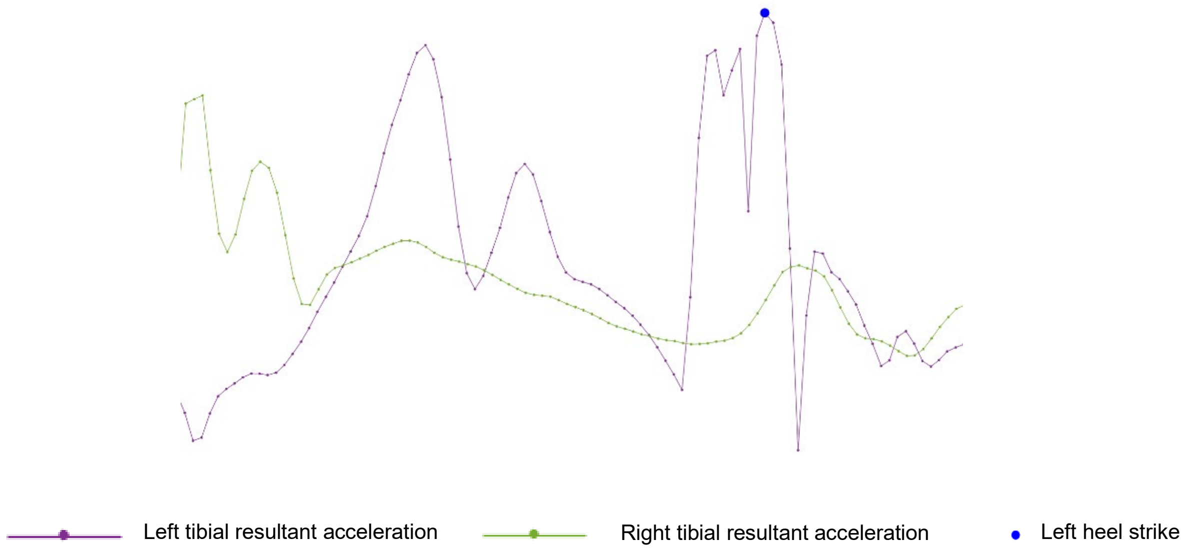

Figure 6.4 Resultant acceleration of a middle-distance runner for one heel strike

It is believed that the double peak in the acceleration trace is a representation of a rearfoot strike, which concurs with the earlier findings of Cavanagh and Lafortune (1980) who demonstrated that the vertical ground reaction force pattern in rearfoot strikers displayed a double peak. Currently, the algorithm determines ground contact for middle-distance runners as the second and highest peak (shown by the blue dot in figure 6.4). This provides a plausible explanation as to why contact time, COM displacement was underestimated, and flight time overestimated compared to the 3D motion analysis system as ground contact had already occurred before it was actually detected in the algorithm. This can be seen in Figure 6.5 which shows the resultant acceleration (surrogate force measure) and corresponding COM trace for the same contact shown in Figure 6.4. The blue dots show the detection of ground contact based on the current algorithm. The red dots indicate where we believe ground contact is occurring.

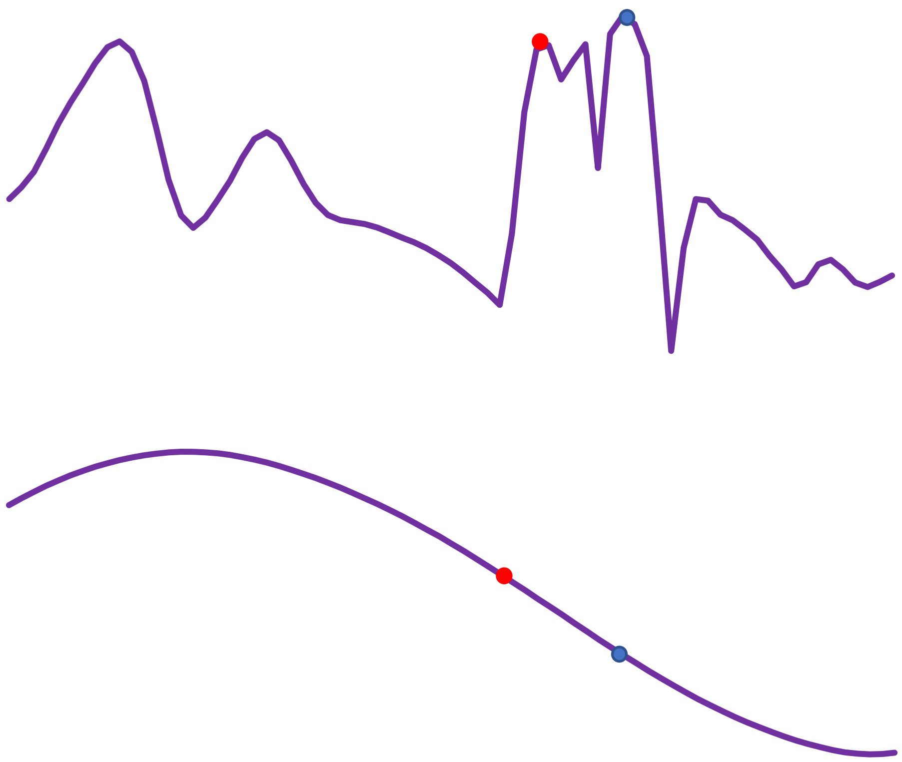

Figure 6.5 Resultant acceleration trace and corresponding COM trace for the same heel strike displayed in Figure 6.4

Looking at the top trace (resultant acceleration), we theorise that the peak with the red dot is where the heel first contacts the ground. As endurance runners tend to land with their toes abducted (pointed outwards) (Cavanagh et al., 1985) we believe that the movement between the red and blue dot is a representation of their greater mediolateral movement during ground contact. From the high-speed video footage, it appears that they ‘roll over’ their foot and therefore we believe that the highest peak (blue dot) should occur once their entire foot has contacted the ground. In terms of COM displacement, it was calculated as the difference between the position of the sensor at heel strike and position of the sensor during mid stance. Based on the COM trace (bottom trace), it is evident that the current position of the sensor at ground contact (blue dot) is detected too close to the minimum point which offers an explanation as to why COM displacement may be currently underestimated. Therefore, it is acknowledged that a limitation of this study is the algorithms ability to detect the correct ground contact point for rearfoot strikers or athletes who display pronation at ground contact. Further investigation and research are needed to determine the precision and accuracy of not only IMeasureU sensors but other inertial measurement units and their ability to correctly identify ground contact based on differences in an individual's gait pattern. While not a primary focus of this study, it would be interesting to know how well inertial measurement units can establish running asymmetry. Currently, few studies have looked at the use of inertial measurement units in detecting running symmetry on a treadmill (Hughes, Jones, Starbuck, Sergeant, & Callaghan, 2019; Kobsar, Osis, Hettinga, & Ferber, 2014; Lee, Sutter, et al., 2010). However, it is thought that overground running on different surfaces may influence how various temporal parameters are detected based on potential inconsistent features in the acceleration trace between individuals and limbs.

6.4.3 Limitations

Given the novel approach of this study, there were a few limitations that warrant discussion. A major limitation of this study was surrounding the technology itself and its current capacity to measure the variables that are needed for each stiffness calculation. While it is understood that technological difficulties are inevitable, the drop out of contacts meant that eight trials couldn't be analysed. In addition, the drop out of contacts didn't allow all five contacts surrounding the force plate contact to be averaged for twenty-eight trials. Although this issue will continue to improve with advancements in this technology, there is still a risk that vital information may be missed regarding an athlete's load if a valid and reliable stiffness measure is found in the near future. In addition, multiple steps are required to export the data from the IMeasureU sensors. Given the large data sets, it is currently not feasible to obtain a real time stiffness measure. However, the recent release of the new Blue Trident IMeasureU sensors (Released June 2019) are believed to deliver real-time metrics and have a faster downloading process. This new release may provide an option for sport practitioners to make immediate decisions and interventions. Finally, while improvements have been made to the algorithm from the initial data collection (Appendix F), further investigation surrounding the position of the IMeasureU sensors, and refinement of the algorithm to better detect an individual's gait/foot striking pattern are needed. With such changes it may be possible to improve our understanding and establish what benefits inertial measurement units hold in determining a field-based measure of lower limb stiffness.

6.4.4 Future Research

Despite the previous section outlining recommendations for imminent research, this section will provide direction for future research based on the idea that a valid and reliable stiffness measure has been found. If refinements to the algorithm and developments of such technology are able to provide a valid and reliable measure of lower limb stiffness in the long-term, future research should focus on more longitudinal measures and the development of a lower-limb stiffness injury model. By being able to examine how training effects an individual's level of stiffness on a more regular basis, it is hoped that researchers/sport practitioners will become closer to determining what that optimal level is for performance enhancement and injury prevention.

6.5 FINAL REMARKS

To date, studies examining longitudinal stiffness measures in athletic populations have so far been isolated to laboratory-based settings. With advancements in technology there is now a possibility for lower limb stiffness to be monitored in the daily training environment. Although a valid and reliable measure of lower limb stiffness still needs to be found, it is hoped that this technology in the future may allow for more frequent sport specific measures to be put in place. It is anticipated that this project is the first step in developing a daily monitoring tool which may provide a proactive approach in managing an athlete's response to load, ultimately reducing their risk of injury. However, further refinements to the algorithm and developments in the IMeasureU sensors are required before this technology can be considered for use in the daily training environment.

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APPENDICES

Appendix A: STROBE Criteria

STROBE Statement—checklist of items that should be included in reports of observational studies

	Item No	Recommendation
Title and abstract	1	(*a*) Indicate the study's design with a commonly used term in the title or the abstract
		(*b*) Provide in the abstract an informative and balanced summary of what was done and what was found
Introduction		
Background/rationale	2	Explain the scientific background and rationale for the investigation being reported
Objectives	3	State specific objectives, including any prespecified hypotheses
Methods		
Study design	4	Present key elements of study design early in the paper
Setting	5	Describe the setting, locations, and relevant dates, including periods of recruitment, exposure, follow-up, and data collection
Participants	6	(*a*) *Cohort study*—Give the eligibility criteria, and the sources and methods of selection of participants. Describe methods of follow-up *Case-control study*—Give the eligibility criteria, and the sources and methods of case ascertainment and control selection. Give the rationale for the choice of cases and controls *Cross-sectional study*—Give the eligibility criteria, and the sources and methods of selection of participants
		(*b*) *Cohort study*—For matched studies, give matching criteria and number of exposed and unexposed *Case-control study*—For matched studies, give matching criteria and the number of controls per case
Variables	7	Clearly define all outcomes, exposures, predictors, potential confounders, and effect modifiers. Give diagnostic criteria, if applicable
Data sources/ measurement	8*	For each variable of interest, give sources of data and details of methods of assessment (measurement). Describe comparability of assessment methods if there is more than one group
Bias	9	Describe any efforts to address potential sources of bias

Study size	10	Explain how the study size was arrived at
Quantitative variables	11	Explain how quantitative variables were handled in the analyses. If applicable, describe which groupings were chosen and why
Statistical methods	12	(*a*) Describe all statistical methods, including those used to control for confounding
		(*b*) Describe any methods used to examine subgroups and interactions
		(*c*) Explain how missing data were addressed
		(*d*) *Cohort study*—If applicable, explain how loss to follow-up was addressed *Case-control study*—If applicable, explain how matching of cases and controls was addressed *Cross-sectional study*—If applicable, describe analytical methods taking account of sampling strategy
		(*e*) Describe any sensitivity analyses
Results		
Participants	13*	(a) Report numbers of individuals at each stage of study—e.g. numbers potentially eligible, examined for eligibility, confirmed eligible, included in the study, completing follow-up, and analysed
		(b) Give reasons for non-participation at each stage
		(c) Consider use of a flow diagram
Descriptive data	14*	(a) Give characteristics of study participants (e.g. demographic, clinical, social) and information on exposures and potential confounders
		(b) Indicate number of participants with missing data for each variable of interest
		(c) *Cohort study*—Summarise follow-up time (e.g., average and total amount)
Outcome data	15*	*Cohort study*—Report numbers of outcome events or summary measures over time
		Case-control study—Report numbers in each exposure category, or summary measures of exposure
		Cross-sectional study—Report numbers of outcome events or summary measures
Main results	16	(*a*) Give unadjusted estimates and, if applicable, confounder-adjusted estimates and their precision (e.g., 95% confidence interval). Make clear which confounders were adjusted for and why they were included
		(*b*) Report category boundaries when continuous variables were categorized

		(*c*) If relevant, consider translating estimates of relative risk into absolute risk for a meaningful time period
Other analyses	17	Report other analyses done—e.g. analyses of subgroups and interactions, and sensitivity analyses
Discussion		
Key results	18	Summarise key results with reference to study objectives
Limitations	19	Discuss limitations of the study, taking into account sources of potential bias or imprecision. Discuss both direction and magnitude of any potential bias
Interpretation	20	Give a cautious overall interpretation of results considering objectives, limitations, multiplicity of analyses, results from similar studies, and other relevant evidence
Generalisability	21	Discuss the generalisability (external validity) of the study results
Other information		
Funding	22	Give the source of funding and the role of the funders for the present study and, if applicable, for the original study on which the present article is based

*Give information separately for cases and controls in case-control studies and, if applicable, for exposed and unexposed groups in cohort and cross-sectional studies.

Note: An Explanation and Elaboration article discusses each checklist item and gives methodological background and published examples of transparent reporting. The STROBE checklist is best used in conjunction with this article (freely available on the Web sites of PLoS Medicine at http://www.plosmedicine.org/, Annals of Internal Medicine at http://www.annals.org/, and Epidemiology at http://www.epidem.com/). Information on the STROBE Initiative is available at www.strobe-statement.org.

Appendix B: Ethics Approval Confirmation Email

Dear Applicant,

Principal Investigator: A/Prof David Greene
Co-Investigator: Dr Mark Moresi, Emma Millett, Dr Elissa Phillips
Student Researcher: Taylor Wileman (HDR Student)
Ethics Register Number: 2016-284H
Project Title: Mechanical loading, injury and performance in high level track and field athletes: A novel approach to athlete monitoring
Risk Level: Low Risk
Date Approved: 16/01/2017
Ethics Clearance End Date: 30/06/2020

This email is to advise that your application has been approved by the Australian Catholic University's Human Research Ethics Committee and confirmed as meeting the requirements of the National Statement on Ethical Conduct in Human Research. This approval will be ratified at the next available meeting and is subject to the following:

- Satisfactory validation of Working with Children Checks;
- Receipt of outstanding permission letters/other approvals;
- Ratification of any outstanding items (e.g. interview/survey questions).

You will be contacted should the Committee raise any issues in relation to the above matters.

Failure to provide outstanding documents to the ACU HREC before data collection commences is in breach of the National Statement on Ethical Conduct in Human Research and the Australian Code for the Responsible Conduct of Research. ACU HREC approval is only valid as long as approved procedures are followed.

Researchers who fail to submit a progress report may have their ethical clearance revoked and/or the ethical clearances of other projects suspended. When your project has been completed a progress/final report form must be submitted. The information researchers provide on the security of records, compliance with approval consent procedures and documentation and responses to special conditions is reported to the NHMRC on an annual basis. In accordance with NHMRC the ACU HREC may undertake annual audits of any projects considered to be of more than low risk.

Clinical Trials - Researchers should refer to the Australian New Zealand Clinical Trials Registry http://www.anzctr.org.au/) for information.

It is the Principal Investigators / Supervisors responsibility to ensure that:
1. All serious and unexpected adverse events (or any matter that might affect the ethical acceptability of the protocol) should be reported to the HREC with 72 hours.
2. Any changes to the protocol must be reviewed by the HREC by submitting a Modification/Change to Protocol Form prior to the research commencing or continuing. http://research.acu.edu.au/researcher-support/integrity-and-ethics/
3. Progress reports are to be submitted on an annual basis. http://research.acu.edu.au/researcher-support/integrity-and-ethics/
4. Protocols can be extended for a maximum of five (5) years after which a new application must be submitted. (The five year limit on renewal of approvals allows the Committee to fully re-review research in an environment where legislation, guidelines and requirements are continually changing, for example, new child protection and privacy laws).

Please do not hesitate to contact the office if you have any queries.

Kind regards,

Kylie Pashley
on behalf of ACU HREC Chair, Dr Nadia Crittenden

Ethics Officer | Research Services
Office of the Deputy Vice Chancellor (Research) Australian Catholic University

Appendix C: Participant & Parent/Guardian Information Letter

PARTICIPANT INFORMATION LETTER

PROJECT TITLE: Real Time Stiffness Monitoring for Performance and Injury Prevention in High Level Track and Field Athletes.

PRINCIPAL INVESTIGATOR:
A/Prof David Greene

CO-INVESTIGATORS:
Dr Mark Moresi, Dr Elissa Phillips,
Dr Emma Millett

STUDENT RESEARCHER:
Ms Taylor Wileman

STUDENT'S DEGREE:
Doctor of Philosophy

Dear Participant,

You are invited to participate in the research project described below.

What is the project about?

Interaction of training load and changes to the way an athlete deals with loading (leg stiffness) could provide vital feedback to athletes, coaches, medical and support staff enabling early injury risk identification and performance maximisation. This information provided in real time to key high performance personnel can allow for effective systems to be put into place to ensure athletes reach their potential. The application of inertial sensor technology in the daily training environment has the potential for a proactive approach to the management of athlete performance and injury. The project aims to validate and implement such technology to quantify lower limb stiffness across four event groups (sprints, hurdles, jumps and endurance) and investigate possible links of this to performance and injury.

Who is undertaking the project?

This project is being conducted by A/Prof David Greene, Dr Mark Moresi, Dr Elissa Phillips, Dr Emma Millett and Ms Taylor Wileman with support from Athletics Australia, the Australian Institute of Sport and the New South Wales Institute of Sport. The information collected will form the basis for the degree of Ms Taylor Wileman at the Australian Catholic University.

Are there any risks associated with participating in this project?

Your participation in this project comes with minimal risk to you as the participant. The tasks required of you are similar to what you would normally complete as part of your regular training for your event.

What will I be asked to do?

As a participant, you will be asked to attend one testing session which will require you to participate in the following activities:

- Jumps Testing: including vertical jumps, drop jumps and single and double leg continuous jumps
- Anthropometry: measurements of your height, weight, limb lengths and some joint breadths will be assessed
- Running Analysis: three-dimensional motion analysis will be gathered of you running at approximately race pace. Jumpers and hurdles may also be required to perform some event specific trial.
- Inertial Sensors: Three inertial sensors incorporating three dimensional accelerometers will be placed on both the left and right distal tibias and an approximate centre of mass (lower back) location. They will be worn during the running and jumping analysis session.
- Injury History Questionnaire: participants will also be asked to complete a short questionnaire (5 minutes) on your previous injury history

How much time will the project take?

The testing session will take approximately 2 hours of your time.

What are the benefits of the research project?

As a participant, you will receive a summary biomechanical analysis report following your testing session highlighting your strengths and areas of improvement which will assist in your training and competition preparation.

Can I withdraw from the study?

Participation in this study is completely voluntary. You are not under any obligation to participate. If you agree to participate, you can withdraw from the study at any time without adverse consequences. Participation in the present study will in no way influence any decisions or relationships you may have in regards to your track and field career including selections to state or national squads, state institutes or academies of sport or national teams.

Will anyone else know the results of the project?

It is anticipated that the summary results of the study will be published in appropriate peer-reviewed journals and conferences. Individual athlete data will be treated confidentially and stored securely, and any data presented to a wider audience will be in the form of summary data with individual athletes non-identifiable.

Will I be able to find out the results of the project?

In addition to receiving your own individual camp-based results, it is anticipated that the summary results of the study will be distributed back to coaches and athletes via Athletics Australia and the State Institute and Academy network.

Who do I contact if I have questions about the project?

If you have any questions regarding the project, please feel free to contact Taylor Wileman via email at taylor.wileman@myacu.edu.au

What if I have a complaint or any concerns?
The study has been reviewed by the Human Research Ethics Committee at Australian Catholic University (review number 2016 284H). If you have any complaints or concerns about the conduct of the project, you may write to the Manager of the Human Research Ethics Committee care of the Office of the Deputy Vice Chancellor (Research).

Manager, Ethics
c/o Office of the Deputy Vice Chancellor (Research)
Australian Catholic University
North Sydney Campus
PO Box 968
NORTH SYDNEY, NSW 2059
Ph.: 02 9739 2519
Fax: 02 9739 2870
Email: resethics.manager@acu.edu.au

Any complaint or concern will be treated in confidence and fully investigated. You will be informed of the outcome.

I want to participate! How do I sign up?
If you are happy to participate in the study, sign the attached consent forms (please ensure you date and sign a copy for yourself and the researchers) and bring the forms with you to your scheduled testing session. If you are under 18 years of age, you will need to complete the “Parent/Guardian Consent Form” with the approval of a parent/guardian.

Yours sincerely,

A/Prof David Greene
Dr Mark Moresi
Dr Emma Millett
Dr Elissa Phillips
Ms Taylor Wileman

PARENT/GUARDIAN PARTICIPANT INFORMATION LETTER

PROJECT TITLE: Real Time Stiffness Monitoring for Performance and Injury Prevention in High Level Track and Field Athletes.

PRINCIPAL INVESTIGATOR:
A/Prof David Greene

CO-INVESTIGATORS:
Dr Mark Moresi, Dr Elissa Phillips,
Dr Emma Millett

STUDENT RESEARCHER:
Ms Taylor Wileman

STUDENT'S DEGREE:
Doctor of Philosophy

Dear Participant,

You are invited to participate in the research project described below.

What is the project about?

Interaction of training load and changes to the way an athlete deals with loading (leg stiffness) could provide vital feedback to athletes, coaches, medical and support staff enabling early injury risk identification and performance maximisation. This information provided in real time to key high performance personnel can allow for effective systems to be put into place to ensure athletes reach their potential. The application of inertial sensor technology in the daily training environment has the potential for a proactive approach to the management of athlete performance and injury. The project aims to validate and implement such technology to quantify lower limb stiffness across four event groups (sprints, hurdles, jumps and endurance) and investigate possible links of this to performance and injury.

Who is undertaking the project?

This project is being conducted by A/Prof David Greene, Dr Mark Moresi, Dr Elissa Phillips, Dr Emma Millett and Ms Taylor Wileman with support from Athletics Australia, the Australian Institute of Sport and the New South Wales Institute of Sport. The information collected will form the basis for the degree of Ms Taylor Wileman at the Australian Catholic University.

Are there any risks associated with participating in this project?

Your child's participation in this project comes with minimal risk. The tasks required of your son/daughter are the similar to what they would normally complete as part of their regular training for their particular event.

What will they be asked to do?

Participants will be asked to attend one testing session which will require them to participate in the following activities;

- Jumps Testing: including vertical jumps, drop jumps and single and double leg continuous jumps
- Anthropometry: measurements of their height, weight, limb lengths and some joint breadths will be assessed
- Running Analysis: three-dimensional motion analysis will be gathered of their running at approximately race pace. Jumpers and hurdles may also be required to perform some event specific trials
- Inertial Sensors: Three inertial sensors incorporating three dimensional accelerometers will be placed on both the left and right distal tibias and an approximate centre of mass (lower back) location. They will be worn during the running and jumping analysis session.
- Injury History Questionnaire: participants will also be asked to complete a short questionnaire (5 minutes) on your previous injury history

How much time will the project take?
The testing session will take approximately 2 hours.

What are the benefits of the research project?
As a participant, your son/daughter will receive a summary biomechanical analysis report following the testing session highlighting their individual strengths and areas of improvement which will assist in their training and competition preparation.

Can participants withdraw from the study?
Participation in this study is completely voluntary. Your child will not be under any obligation to participate. If you and your child agree to participate, they are free to withdraw from the study at any time without adverse consequences. Participation in the present study will in no way influence any decisions or relationships they may have in regards to their track and field career including selections to state or national squads, state institutes or academies of sport or national teams.

Will anyone else know the results of the project?
It is anticipated that the summary results of the study will be published in appropriate peer-reviewed journals and conferences. Individual athlete data will be treated confidentially and stored securely, and any data presented to a wider audience will be in the form of summary data with individual athletes non-identifiable.

Will I be able to find out the results of the project?
In addition to receiving your child's own individual camp-based results, it is anticipated that the summary results of the study will be distributed back to coaches and athletes via Athletics Australia and the State Institute and Academy network.

Who do I contact if I have questions about the project?
If you have any questions regarding the project, please feel free to contact Taylor Wileman via email at taylor.wileman@myacu.edu.au

What if I have a complaint or any concerns?
The study has been approved by the Human Research Ethics Committee at Australian Catholic University (approval number 2016 284H). If you have any complaints or concerns about the conduct of the project, you may write to the Chair of the Human Research Ethics Committee care of the Office of the Deputy Vice Chancellor (Research).

Chair, HREC
c/o Office of the Deputy Vice Chancellor (Research)
Australian Catholic University
Melbourne Campus
Locked Bag 4115
FITZROY, VIC, 3065
Ph: 03 9953 3150
Fax: 03 9953 3315
Email: res.ethics@acu.edu.au

Any complaint or concern will be treated in confidence and fully investigated. You will be informed of the outcome.

I am happy for my child to participate! How do we sign up?
If you and your child are happy to participate in the study, sign the attached consent forms (please ensure you data and sign a copy for yourself and the researchers) and bring the forms with you to your child's scheduled testing session. As your child is under 18 years of age, you will need to complete the "Parent/Guardian Consent Form" with both their signature indicating willingness to participate and your approval as their parent/guardian.

Yours sincerely,

A/Prof David Greene
Dr Mark Moresi
Dr Emma Millett
Dr Elissa Phillips
Ms Taylor Wileman

Appendix D: Informed Consent/Assent Forms

CONSENT FORM

Copy for Participant to Keep

TITLE OF PROJECT: Real Time Stiffness Monitoring for Performance and Injury Prevention in High-Level Track and Field Athletes

PRINCIPAL INVESTIGATOR: A/Prof David Greene

CO-INVESTIGATORS: Dr Mark Moresi, Dr Emma Millett, Dr Elissa Phillips

STUDENT RESEARCHER: Ms Taylor Wileman

I.. *(the participant)* have read (*or, where appropriate, have had read to me)* and understood the information provided in the Letter to Participants. Any questions I have asked have been answered to my satisfaction. I agree to participate in the jump and running tasks. I realise that I can withdraw my consent at any time without any adverse consequences and that participation in the present study will in no way affect any selection process for national squads, teams or state institute/academy squads. I agree that research data collected for the study may be published or may be provided to other researchers in a form that does not identify me in any way.

NAME OF PARTICIPANT: ....

SIGNATURE: .... DATE ....

SIGNATURE OF PRINCIPAL INVESTIGATOR: ....

DATE: ………………………..

SIGNATURE OF STUDENT RESEARCHER: ....

DATE: ....………

CONSENT FORM

Copy for Researcher to Keep

TITLE OF PROJECT: Real Time Stiffness Monitoring for Performance and Injury Prevention in High-Level Track and Field Athletes

PRINCIPAL INVESTIGATOR: A/Prof David Greene

CO-INVESTIGATOR: Dr Mark Moresi, Dr Emma Millett, Dr Elissa Phillips

STUDENT RESEARCHER: Ms Taylor Wileman

I... *(the participant)* have read (*or, where appropriate, have had read to me)* and understood the information provided in the Letter to Participants. Any questions I have asked have been answered to my satisfaction. I agree to participate in the jump and running tasks. I realise that I can withdraw my consent at any time without any adverse consequences and that participation in the present study will in no way affect any selection process for national squads, teams or state institute/academy squads. I agree that research data collected for the study may be published or may be provided to other researchers in a form that does not identify me in any way.

NAME OF PARTICIPANT: ....

SIGNATURE: .... DATE: ....

SIGNATURE OF PRINCIPAL INVESTIGATOR: ....
DATE: ………………………..

SIGNATURE OF STUDENT RESEARCHER: ....
DATE: ....……….

PARENT/GUARDIAN CONSENT FORM

Copy for Participant to Keep

TITLE OF PROJECT: Real Time Stiffness Monitoring for Performance and Injury Prevention in High-Level Track and Field Athletes

PRINCIPAL INVESTIGATOR: A/Prof David Greene
CO-INVESTIGATOR: Dr Mark Moresi, Dr Emma Millett, Dr Elissa Phillips
STUDENT RESEARCHER: Ms Taylor Wileman

I .... (*the parent/guardian*) have read (*or, where appropriate*, have had read to me) and understood the information provided in the Letter to the Participants. Any questions I have asked have been answered to my satisfaction. I agree that my child, nominated below, may participate in the jump and running tasks. I realise that I can withdraw my consent at any time without any adverse consequences and that participation in the present study will in no way affect any selection process for national squads, teams or state institute/academy squads. I agree that research data collected for the study may be published or may be provided to other researchers in a form that does not identify my child in any way.

NAME OF PARENT/GUARDIAN: ....

SIGNATURE ....……………………..… DATE: ………

NAME OF CHILD: ....

SIGNATURE OF PRINCIPAL INVESTIGATOR: ……………………………… DATE: ....

SIGNATURE OF STUDENT RESEARCHER:…………………..……………… DATE: ……….

ASSENT OF PARTICIPANTS AGED UNDER 18 YEARS

I …………………….. (*the participant aged under 18 years*) understand what this research project is designed to explore. What I will be asked to do has been explained to me. I agree to take part in the jump and running tasks, realising that I can withdraw at any time without having to give a reason for my decision.

NAME OF PARTICIPANT AGED UNDER 18: ....

SIGNATURE: …………………………………………….. DATE: ………

SIGNATURE OF PRINCIPAL INVESTIGATOR (or SUPERVISOR): …………………....

DATE: ………

SIGNATURE OF STUDENT RESEARCHER (if applicable)]: ....

DATE: ………

PARENT/GUARDIAN CONSENT FORM

Copy for Researcher to Keep

TITLE OF PROJECT: Real Time Stiffness Monitoring for Performance and Injury Prevention in High-Level Track and Field Athletes

PRINCIPAL INVESTIGATOR: A/Prof David Greene
CO-INVESTIGATOR: Dr Mark Moresi, Dr Emma Millett, Dr Elissa Phillips
STUDENT RESEARCHER: Ms Taylor Wileman

I .... (*the parent/guardian*) have read (*or, where appropriate*, have had read to me) and understood the information provided in the Letter to the Participants. Any questions I have asked have been answered to my satisfaction. I agree that my child, nominated below, may participate in the jump and running tasks. I realise that I can withdraw my consent at any time without any adverse consequences and that participation in the present study will in no way affect any selection process for national squads, teams or state institute/academy squads. I agree that research data collected for the study may be published or may be provided to other researchers in a form that does not identify my child in any way.

NAME OF PARENT/GUARDIAN: ....

SIGNATURE ....……………… DATE: ………………

NAME OF CHILD: ....

SIGNATURE OF PRINCIPAL INVESTIGATOR:……………………… DATE: ……………

SIGNATURE OF STUDENT RESEARCHER:…………………………… DATE: ……………

ASSENT OF PARTICIPANTS AGED UNDER 18 YEARS

I ……………………… (*the participant aged under 18 years*) understand what this research project is designed to explore. What I will be asked to do has been explained to me. I agree to take part in the jump and running tasks, realising that I can withdraw at any time without having to give a reason for my decision.

NAME OF PARTICIPANT AGED UNDER 18: ....

SIGNATURE: …………………………………………………………………… DATE: ……………

SIGNATURE OF PRINCIPAL INVESTIGATOR (or SUPERVISOR): …………………..

DATE: …………..

SIGNATURE OF STUDENT RESEARCHER (if applicable)]: ....

DATE: ……………

Appendix E: Screening Form

AUSTRALIAN CATHOLIC UNIVERSITY

SCREENING FORM

Real Time Stiffness Monitoring for Performance and Injury Prevention in High Level Track and Field Athletes

Name: ....

Date of Birth: .... Age: ....

Event Group: ……………………………………… PBs: ……………………

Gender: Female / Male

Height: ....

Weight: ....

Left Leg Length: ....

Left Knee Width: ....

Left Ankle Width: ....

Left Shoulder Offset: ....

Left Elbow Width: ....

Left Wrist Width: ....

Left Hand Thickness: ....

Right Leg Length: ....

Right Knee Width: ....

Right Ankle Width: ....

Right Shoulder Offset: ....

Right Elbow Width: ....

Right Wrist Width: ....

Right Hand Thickness: ....

Leg Dominance: ....

Exercise History:

Athletic Background: ....

Highest Athletic Level:

....

Training Years:

....

Average Training Hours Per Week: ....

How Often Per Week: ....

Typical Training Week:

....

....

....

Predominately What Type of Exercise:

....

Sports Injury Details:

Please List Any Current or Recurring Injuries:

....

....

....

Please List Any Previous Injuries Within the Last 2 Years. When and How They Occurred:

....

....

....

....

Do you suffer any recurring pain in any joint when participating in sport?

Yes No

Do you require specific taping/padding for a previous injury?

Yes No

If Yes, Please Detail: ....

Appendix F: Improvements to Algorithm Based on Initial Data Collection

This section will outline the initial data collection and highlight improvements to the IMeasureU sensor software and algorithm based on these findings. Our initial validation attempt involved the collection of 150 sprints. However, due to technological issues such as the IMeasureU sensors not capturing the data only 18% of the data could be analysed. This data collection provided a foundation for improvements to the IMeasureU sensor software and algorithm. The three main developments and improvements made prior to the present data collection are outlined below.

1. COM Dampening Effect

The original plan of this research was to only use data from the IMeasureU sensor placed on an approximate COM location. However, the results from the initial data collection showed that there was a dampening effect. The correlation results for contact time identified a weak relationship between the 3D motion analysis system and the IMeasureU sensor located on the COM (r = 0.2). A weak correlation was evident as we were unable to consistently identify a feature in the acceleration trace that correlated with ground contact time. This was believed to be the result of dampening which concurred with the earlier findings of Bergamini et al. (2012). Therefore, we decided to use the tibia sensors in combination with the COM sensor to detect ground contact time. We were able to better identify the heel strike phase with the tibia sensor and as a result a moderate correlation was evident (r = 0.7). The need for all three IMeasureU sensors were identified.

2. Manual Sync

The use of all three IMeasureU sensors presented a challenge earlier on as they had to be manually synced together. However, further pilot testing led to the development of software which synced the IMeasureU units. More recently, the IMeasureU-Vicon sync function was released with the new Vicon Nexus software. However, given the need for a higher sample frequency, the IMeasureU-Vicon sync feature couldn't be used. Therefore, the IMeasureU sensors were synced through the MATLAB algorithm and each step was matched with Vicon through the use of high-speed video footage.

3. Saturation of Force

Another issue found was surrounding the use of the peak acceleration from the tibia sensors as our surrogate force measure. However, during the analysis of data, it was evident that the IMeasureU sensors were 'maxing out' at a certain point. When we had a closer look at the saturation of the tibia sensors, it was evident that the sensors were saturating mainly in the x-axis, which aligned with the direction of the runner. Therefore, we decided to use the resultant peak acceleration as the surrogate force measure. Also, it was believed that the values estimated from the individual saturated axis would be further from the true value. It was decided that by using the resultant it would still be a comparable measure of loading of the leg (i.e. more saturate = larger resultant = larger load).