Modifiable Risk Factors of Hamstring Strain Injury: Assessment, Performance and Training

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List of abbreviations

- **BF_{LH}** – Bicep femoris long head
- **BF_{SH}** – Bicep femoris short head
- **SM** – Semimembranosus
- **ST** – Semitendinosus
- **MTU** – Muscle-tendon unit
- **HSI** – Hamstring strain injury
- **FL** – Fascicle length
- **AA** – Aponeurosis angle
- **PA** – Pennation angle
- **MT** – Muscle thickness
- **EMG** – Electromyography
- **MVIC** – Maximum voluntary isometric contraction
- **US** – Ultrasound
- **MRI** – Magnetic resonance imaging
- **FOV** – Field of view
- **NHE** – Nordic hamstring exercise
- **GHR** – Glute-hamstring raise
- **IMTP** – Isometric mid-thigh pull
- **CMJ** – Countermovement jump
- **DOMS** – Delayed onset muscle soreness
- **PRISMA** - *Preferred Reporting Items for Systematic Reviews and Meta-Analyses*
- **1D,2D or 3D** – One-, two- or three-dimensional
- **TO** – Take off
- **TD** – Touch down

Glossary of terms

Chapter 5 - Study 3 - Part A & B

- **Peak knee flexion/extension** – Peak knee flexion/extension angle (full knee extension = 0°)
- **Peak hip flexion/extension** – Peak hip flexion/extension angle (neutral extension = 0°)
- **Change in knee angular velocity** – Difference in angular velocity values between time-points

Chapter 6 - Study 4

- **Knee angle at break point** – knee angle identified at 20 deg/s of knee extension velocity (full knee extension = 180°)
- **Knee angle at break point relative to the horizontal** – Difference between the knee angle at break point and the horizontal plane
- **Change in knee angle** – Difference between the starting knee angle and the knee angle at break point and the horizontal plane

List of Publications and Presentations

- Sprint vs Nordic hamstring exercise: Effect of initial sprint ability on the magnitude of adaptations to eccentric hamstring strength and Bicep femoris fascicle length. **(ALTIS Virtual Apprentice Coach Program, Online September 2020)**
- Eccentric hamstring strength and sagittal plane lower limb running kinematics across team sports. **(UKSCA Conference Podium presentation, Online September 2020)**
- The impact of hamstring strain injury risk factors on submaximal running kinematics and hamstring activation. **(European Congress of Sport Sciences (ECSS) Conference Poster presentation, Online October 2020)**
- The effects of a seven-week sprint vs Nordic training intervention on the modifiable risk factors of hamstring strain injury and performance. **(National Strength and Conditioning Association (NSCA) National Conference Podium presentation, July 2020)**
- Retention of adaptations to eccentric hamstring strength and bicep femoris fascicle length from a seven-week sprint or Nordic training intervention. **(NSCA National Conference Poster presentation, July 2020)**
- Effect of the Nordic hamstring exercise ability on in-vivo fascicle dynamics during variations of the Nordic hamstring exercise**. (ISBS Conference Poster presentation, Online 2020)**
- The Effect of Nordic Hamstring Exercise Intervention Volume on Eccentric Strength and Muscle Architecture Adaptations: A Systematic Review and Meta-analyses, Corrections and Rely to. **(Co-author, Sports Medicine, 2019)**
- A Systematic Review of Surface Electromyography Onset Activation Analysis Techniques During Running Tasks. **(British Association of Sports and Exercise Sciences (BASES) Biomechanics Interest Group Poster presentation, University of Salford 2017)**
- Effect of Filtering Window Durations on Peak and Mean Electromyography Amplitude of The Bicep Femoris During the Glute-Ham Raise Exercise. **(NSCA National Conference Poster presentation, Indianapolis 2017)**
- Effect of Different Onset Thresholds on Electromyography Variables of The Bicep Femoris During the Glute-Ham Raise Exercise. **(NSCA National Conference Poster presentation, Indianapolis 2017)**

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Abstract

Hamstring strain injuries (HSIs), which are typically sustained during high velocity movements (i.e., sprinting or high velocity running), commonly involves the bicep femoris long head (BF_{LH}). To date, the Nordic hamstring exercise (NHE) is the most effective training modality for reducing HSIs within team sports, however, compliance to training prescription is commonly low which can influence the effectiveness of the intervention. Alternative training modalities have been advocated within the literature as being effective at decreasing the risk of sustaining a future HSI by increased eccentric hamstring strength and BF_{LH} fascicle length (FL). There are several alternative training modalities, including hip dominant exercises and sprinting, which are relatively unexplored within the literature, with between just one and two previous investigations of these modalities, respectively. Additionally, previous studies commonly lack ecological validity, whereby the use of a single exercise would be unusual within sport practice. The overarching aim of this thesis was to observe the HSI modifiable risk factors within a cyclical-practice approach, which could be used within practice, this includes measurement and identification, identifying their influence on performance and observing the effect of known exercises and training. With specific objectives which will aim to answer gaps within the literature, the objectives were to firstly examine the methods used to assess BF_{LH} FL (Chapter 3 and 4). Secondly, the thesis investigated potential mechanisms of HSIs during running when accounting for eccentric hamstring strength and BF_{LH} FL (Chapter 5). Thirdly, the thesis identified characteristics such as kinematics, electromyography (EMG) and *in-vivo* dynamics of the NHE and variations (Chapter 6). Finally, the thesis observed the effect of a short-term (seven-week), lower-limb resistance training intervention including a hip dominant exercise, with the addition of the either the NHE or sprinting, on eccentric hamstring strength, BF_{LH} FL and measures of athletic performance (Chapter 7).

There is a potential for large degree of error in the measurement of BF_{LH} FL, therefore, investigations were required to determine the most appropriate estimation equation (if required) and imaging field of view (FOV). Therefore, the focus of Chapter 3 and 4, was to determine the most reliable, accurate and low-cost methods of assessing BF_{LH} FL using the ultrasound (US). Between three commonly used methods to estimate BF_{LH} FL, Chapter 3 demonstrates trivial differences between all methods, in 13 male team sport athletes (left and right BF_{LH} n = 26), although the methods cannot be used interchangeably. Chapter 4 aimed to identify if any differences exist between different imaging FOV (6-cm vs. 10-cm) within 16 male team sport athletes (left and right BF_{LH} n = 32). Trivial to small differences were noted between 6-cm and 10-cm FOV, however, there was minimal agreement and incongruency between the two FOV. Although, one estimation equation presented congruent differences between FOV, with the 6-cm FOV consistently overestimating BF_{LH}.

In Chapter 5, physically active males (n = 18) were grouped (high- and low-risk) by relative measures of eccentric hamstring strength and BF_{LH} FL, due to nearly perfect associations between the two measures observed within the present thesis. Peak and waveform kinematic and hamstring EMG characteristics were observed during treadmill running at various running velocities (8-16 km·hr^{-1}). Significant and meaningful differences were observed between the two groups for peak and waveform knee and hip kinematics and BF_{LH} EMG measurements. The differences observed in Chapter 5 could be defined as being unfavourable, especially as the differences occur both within the take off and late swing phases, as being defined as areas

of interest for the incidence of HSI, with alterations in pelvic control and velocity and activation of the hamstrings under lengthening conditions.

The NHE is one of the most extensively researched hamstring exercises, however, to date it is currently unknown what is occurring within the muscle during the exercise and potential variations. Within Chapter 6, kinematic, EMG and *in-vivo* ultrasound imaging was performed, while 13 participants (10 males, 3 females) with resistance training experience, performed NHE variations (0°, -20° and +20° horizontal plane performance angles). There were significant and meaningful differences in kinematic, neuromuscular and BF_{LH} *in vivo* muscle mechanics, with changes in NHE performance angle manipulating the lever arm through which the centre of mass from the knee up is acting. Differences in instantaneous knee angle, MTU length and neuromuscular contributions of the BF_{LH} and the ST to the task at break point could be related to the altered starting position of each performance angle. There were meaningful differences identified within the early-mid range of movement (0-40% time normalized to break point between variations), where greater fascicle shortening was observed within the decline and flat NHE variations. This is potentially explained by the contractile components within the BF_{LH}, to take up more slack which would be present within the elastic component (i.e., distal tendon), which is under less strain within the early stages of the movement. Whereas there were no likely or meaningful differences identified between variations at the mid-end range (40-100% time). Therefore, all variations result in a similar magnitude of relative fascicle lengthening, which may indicate that similar positive adaptations in eccentric hamstring strength and BF_{LH} FL would be attained from there utilisation.

Finally, seven-weeks of lower limb resistance training alone (control n = 10), incorporating a hip dominant exercise alone, results in significant and meaningful increases in eccentric hamstring strength and BF_{LH} FL. However, both the addition of either the low volumes of NHE (n = 15, 2 x 4 with progressive intensity, twice per week) or sprinting (n = 13, 200-350 m per week) resulted in a greater magnitude of increase in eccentric hamstring strength and BF_{LH} FL. The NHE was the more effective at increasing both eccentric hamstring strength and BF_{LH} FL than sprinting, however, the multi-modal approach (NHE or sprinting plus a hip dominant based exercise) as used within the present thesis and would be more frequent in elite practice, is superiorly effective than using the single modality alone. Additionally, and unsurprisingly, a seven-week lower-limb resistance training program was effective at increasing measures of athletic performance (sprint, jump and isometric mid-thigh pull), while simultaneously only inducing low-moderate delayed onset muscle soreness.

1 Introduction

1.1 Statement of Originality of Research

The work contained in this thesis is to the best of my knowledge and belief; original, having not been published previously or written by another person except where due reference is made. The body of research contained within this thesis highlighted running characteristic differences between those at a high- and low-risk of future HSI. Additionally, the *in-vivo* observations of the bicep femoris (BF) during exercise are first of their kind. Finally, the training intervention performed as part of the present body of research, observed the effect of an ecologically valid training program on measures of hamstrings, specifically associated with HSI risk. It is anticipated that the results yielded from this body of research will, therefore be of high impact and thus great benefit to sport performance researchers and practitioners alike.

1.2 Alarming statistics and current research

Muscle strain injuries, such as HSI, are extremely prevalent within sports, particularly where high-speed movement is essential for sporting actions. HSIs represent the most frequent injury incidence, with considerably high related costs estimated €500,000 per injury within elite European football and approximately 15 - 20 missed matches per injury across sports (Brooks, Fuller, Kemp, & Reddin, 2005, 2006; Ekstrand, Hagglund, & Walden, 2011; Ekstrand, Walden, & Hagglund, 2016; Ernlund and Vieira, 2017; Fuller, Taylor, Douglas, & Raftery, 2020; Woods et al., 2004). While high rates of any specific injury are alarming, what is most concerning is the high incidence of recurrence of HSI, with many of these recurrent injuries typically being of greater severity than any initial HSI (Brukner, Nealon, Morgan, Burgess, & Dunn, 2014; Croisier, Forthomme, Namurois, Vanderthommen, & Crielaard, 2002; Opar, Williams, & Shield, 2012; van der Horst, Backx, Goedhart, Huisstede, & Group, 2017; Wangensteen et al., 2016). Globally, multi-sport longitudinal data (including; elite Australian Football League (AFL), rugby union, soccer, athletics, major and minor league baseball, National Football League (NFL)) demonstrates that the occurrence of HSIs has been gradually increasing for a number of years potentially as a result of increased focus of training, accuracy of reporting and increased playing and training intensities and volumes, although the most recent report within AFL does indicate a reduction in HSI incidences (Askling, Tengvar, Tarassova, & Thorstensson, 2014; Brooks, et al., 2006; Camp et al., 2018; Ekstrand, Walden, et al., 2016; Gabbe, Bennell, & Finch, 2006; Mack et al., 2020; Opar et al., 2014; Orchard, Farhart, Kountouris, James, & Portus, 2010; Roe, Murphy, Gissane, & Blake, 2018; Ruddy et al., 2017).

The high financial and performance burden along with the high initial and recurrent rates of HSI incidence make understanding potential mechanisms and developing processes, both with regards to assessment and training, a crucial area of research for both researchers and practitioners alike. To date, previous research has been able to establish and postulate a number of potential risk factors that contribute to HSI incidence (Bahr, 2016; Fousekis, Tsepis, Poulmedis, Athanasopoulos, & Vagenas, 2011; Opar et al., 2015; Opar, et al., 2012; Orchard, 2001), with some attempts made at injury prediction modelling (Ruddy, et al., 2017). Researchers have also identified a series of both laboratory and field-based approaches to hamstring muscle assessment, such as strength, architecture and activation – typically from a readiness or preparedness perspective (Chalker, Shield, Opar, & Keogh, 2016; Opar, Piatkowski, Williams, & Shield, 2013; Timmins, Shield, Williams, Lorenzen, & Opar, 2015; Timmins, Shield, Williams, & Opar, 2016; Wollin, Purdam, & Drew, 2016; Wollin, Thorborg, & Pizzari, 2018). Further research has also proposed a number of potential training methods that could be used to influence HSI incidence rates within sports, with varying degrees of success (Goode et al., 2015; Thorborg et al., 2017; van Dyk, Behan, & Whiteley, 2019). Although, training intervention studies have more frequently observed the effect on HSI risk factors such as strength and muscle architecture inferring the potential for these interventions to have similar effects on HSI incidence rates, without observing HSI occurrence, due to the complex and time-consuming nature of HSI surveillance within sport.

1.3 Running

High speed running is by far the most frequently reported action where HSIs within sport (Ertelt and Gronwald, 2017; Malone et al., 2018; Shield and Murphy, 2018; Van Hooren and Bosch, 2017a, 2017b, 2018). It is thought that muscle strains are likely to occur when the muscle is subjected to high tensile loads while under high levels of activation, although there is also some research interest in the role of neuromuscular inhibition in HSI incidence (Blandford, Theis, Charvet, & Mahaffey, 2018; Bourne, Opar, Williams, Al Najjar, & Shield, 2016; Fyfe, Opar, Williams, & Shield, 2013). Within the running gait cycle, these determinants of strain injuries are found to occur during the late swing and early stance phase; hence the time point for the incident of a HSI within high-speed running has been described to occur during these phases, with two observational case studies proposing the onset of injury could have occurred during the late swing phase (Heiderscheit et al., 2005; Schache, Wrigley, Baker, & Pandy, 2009). Although, there is also some suggestion that HSIs could occur during the early stance phase of running, due to the evidence of a high external extension moment as the knee passes over the foot (Van Hooren and Bosch, 2017a, 2017b, 2018). The exact occurrence of HSIs is an ongoing debate within the literature, with more research required during different phases of running – however, it has been suggested that this may simply be a "academic argument" that will be difficult to conclude (Pizzari, Green, & van Dyk, 2020).

The bicep femoris long head (BF_{LH}) has been identified as the most commonly afflicted hamstring muscle during HSI events (Chumanov, Heiderscheit, & Thelen, 2011; Ertelt and Gronwald, 2017; Evangelidis et al., 2016; Heiderscheit, et al., 2005; Schache, Dorn, Blanch, Brown, & Pandy, 2012; Woods, et al., 2004). One potential explanation for this could be that the peak stretch of the BF_{LH} muscle-tendon unit (MTU) occurs during the terminal swing phase of running, commonly exceeding the muscle's resting length by up to 12% (Dolman, Verrall, & Reid, 2014), leading to greater eccentric load. Despite the tendon component potentially influencing the magnitude of MTU stretch, the composition of the BF_{LH} MTU is predominantly muscle or contractile elements (Kellis, Galanis, Kapetanos, & Natsis, 2012; Kellis, Galanis, Natsis, & Kapetanos, 2010); highlighting that the muscular component of the BF_{LH} should be at the forefront for any hamstring assessment, more specifically individual muscle architecture assessment, as strength assessments will assess the muscle group entirely not individually.

1.4 Hamstring training

Current research is fairly conclusive on the positive effect of the Nordic hamstring exercise (NHE), a supramaximal eccentric exercise, on the reduction of HSIs in addition to the associated risk factors of HSIs (Bourne et al., 2018; Cuthbert et al., 2019; van Dyk, et al., 2019). Both the high frequency of HSI within sport and a trend of elite sporting environments continually not adopting the suggested eccentric strength training (Ekstrand, Walden, et al., 2016; McCall, Dupont, & Ekstrand, 2016), validates the need for further exploration. This is to provide a greater awareness and understanding of what methodologies could be utilized by practitioners to mitigate the risk of HSI incidence. The implementation of eccentric based training methods (e.g. NHE, fly wheel, isokinetics) has been the most frequently researched (Cuthbert, et al., 2019). However, currently there is very little understanding of what is actually occurring at the fascicle level within the hamstring, particularly during eccentric exercises (Van Hooren and Bosch, 2017a, 2018). This is due to very few studies having investigated the *in-vivo* dynamics of the muscle fascicles during hamstring exercises (Cataneo, 2018). The potential benefit of this information would be to aid practitioners with programme design, specifically by helping them to achieve the positive adaptations that could mitigate the risk factors of HSI.

Along with the continual non-adoption of eccentric hamstring training practices by coaches; frequently the compliance of athletes to eccentric hamstring training is also something to be desired (Bahr, Thorborg, & Ekstrand, 2015; Bourne, et al., 2018; Gabbe, Branson, & Bennell, 2006; McCall, et al., 2016; Shield and Bourne, 2018; van der Horst, Smits, Petersen, Goedhart, & Backx, 2015). Despite a minimum effective compliance or minimum effective dose currently being unknown, it could be suggested that an effective compliance or dose is not only one that has training validity (i.e., establishment of effects as a result of training), but also ecological validity (i.e., easily utilised within practice, without interfering with other training due to fatigue or time). Therefore, current research requires expanding into hamstring training which can be embedded into elite practice while considering effects on HSI risk factors or incidence.

1.5 Overarching Aim of the Thesis

The overarching aim of the present thesis was to inform the utilisation of a cyclical-practice format (assessment, performance, and training). This includes establishing methods to effectively assess the primary modifiable risk factors, specifically BF_{LH} architecture, identifying how the primary modifiable risk factors influence running characteristics (kinematics and EMG) and subsequently how different forms of training can potentially mitigate HSI risk via favourable adaptations to the primary modifiable risk factors, with the use of a high-risk task such as sprint running. Informed practise would typically take a cyclical system-based approach to this global aim making it relevant to the current thesis, where they look at appropriate methods of assessment (i.e., how to accurately measure the modifiable risk factors), followed by determining how the measurements may influence athletic performance and finally establishing the most effective training method. To make the conclusions of the thesis as applicable as possible, the assessments need to be efficient and effective, the task needed to be relevant to high-risk activities and the training needed to be ecologically valid.

2 Literature Review

2.1 Chapter overview

The literature review includes a detailed overview of the current research methods and data surrounding the mechanisms (2.2) and occurrence of HSIs within sport (2.3), the risk factors associated with HSIs (2.4), current practices of hamstring training (2.5, 2.6 and 2.7) and finally the measurement of the hamstrings during rest, static and dynamic tasks (2.8 and 2.9).

2.2 Mechanisms behind injury during running

High velocity running for athletes has been identified as a task having an elevated risk of HSI incidence, with the most hazardous portion being the terminal swing phase of the running gait (Higashihara, Nagano, Ono, & Fukubayashi, 2016; Kenneally-Dabrowski, Brown, Warmenhoven, et al., 2019; Nagano, Higashihara, Takahashi, & Fukubayashi, 2014). Two case studies have been published, which identify the approximate time occurrence of a HSI-type event during running (Heiderscheit, et al., 2005; Schache, et al., 2009), with both supporting the previous indications of HSI occurrence during the terminal swing phase. The authors of both case studies identified key deviations in; maker trajectories, velocity, joint angles (trunk, hip and knee), MTU lengths, joint torques and contralateral ground reaction force deviations (Heiderscheit, et al., 2005; Schache, et al., 2009). These deviations were proposed to be events pre- and post HSI occurrence, concluding that the HSI event occurred during the terminal swing phase (Heiderscheit, et al., 2005; Schache, et al., 2009). Despite this suggestion, biomechanical deviations occurred within both the stance and swing phases (Kenneally-Dabrowski, Brown, Lai, et al., 2019), making it difficult to correctly identify an exact estimation of injury occurrence (i.e. during the terminal swing phase or stance phase), as the deviations could therefore be a cause or result of injury (i.e. biomechanical deviation leading to injury or biomechanical deviation as a result of injury) (Heiderscheit, et al., 2005; Kenneally-Dabrowski, Brown, Lai, et al., 2019; Schache, et al., 2009).

One possible hypothesis as to why a HSI event would be more likely to occur during the swing phase, is during this time the hamstring MTU reaches its maximum length (>100% of anatomical resting length) at the greatest lengthening velocities leading to greater muscle tension and eccentric loading (Thelen, Chumanov, Best, Swanson, & Heiderscheit, 2005; Thelen et al., 2005). Furthermore, of the three biarticular hamstring muscles, the BF_{LH} undergoes the greatest stretch, in contrast to both the semimembranosus (SM) and semitendinosus (ST) (Dolman, et al., 2014). Dolman and colleagues (2014), estimated that due to the increased stretch that is experienced by the BF_{LH}, this single muscle is required to exert a greater force in order to resist knee extension and hip flexion than both the SM and ST. Although caution should be taken with this conclusion, as their calculation of force was based on the change in length of the hamstrings, applying the laws of elastic spring motion (force produced is proportional to displacement) (Dolman, et al., 2014), which is unlikely within muscular components of the hamstrings, specifically the BF_{LH} and SM which have an increased membranous composite (Kellis, et al., 2012). Contrastingly, the greatest lengthening velocities have been shown to occur within the ST (Chumanov, et al., 2011; Thelen, Chumanov, Best, et al., 2005; Thelen, Chumanov, Hoerth, et al., 2005), which could be a result of the increased tendinous composition of the ST permitting greater functioning velocities

(Kellis, et al., 2012). Therefore, the synchronization of the high magnitude of muscle lengthening, in addition to high velocities occurring during the terminal swing phase could therefore be contributing to the high incidence of HSI during running tasks, specifically for the BF_{LH}, which has been identified as the most commonly injured hamstring muscle, within HSI events (Connell et al., 2004; Koulouris and Connell, 2003, 2005; Koulouris, Connell, Brukner, & Schneider-Kolsky, 2007).

In order to resist knee extension and hip flexion, the hamstrings are required to perform high velocity and high force action; which has been described as being of an eccentric nature (i.e. active/resisted fascicle lengthening) (Chumanov, et al., 2011; Thelen, Chumanov, Best, et al., 2005; Thelen, Chumanov, Hoerth, et al., 2005). Although this notion of an eccentric muscle action used to resist the terminal knee extension, has recently been questioned with researchers describing it to be an isometric action (Van Hooren and Bosch, 2017a, 2017b, 2018). Suggesting that much of the lengthening that is occurring, is within the passive components (e.g., tendon), whereas the active components are bearing in a constant isometric-like state (Van Hooren and Bosch, 2017a, 2017b, 2018). However, regardless of what actions are occurring to the individual components, the total MTU undergoes lengthening action to a greater degree than their resting length (Chumanov, et al., 2011; Thelen, Chumanov, Best, et al., 2005; Thelen, Chumanov, Hoerth, et al., 2005), furthermore HSI incidence is still occurring at high rates in many sports. This highlights two key considerations; firstly, the degree to which the hamstrings lengthen, and the velocity of lengthening will increase in a linear fashion with running velocity, with both factors continuing to contribute to the high rate of HSIs particularly during high speed running (Chumanov, et al., 2011; Thelen, Chumanov, Best, et al., 2005; Thelen, Chumanov, Hoerth, et al., 2005). Secondly, does the question of eccentric vs isometric action truly matter, the answer here is no – as the hamstrings ability to produce force is a key factor in the rate HSI incidence (Bourne, Opar, Williams, & Shield, 2015; Bourne, et al., 2018; Green, Bourne, & Pizzari, 2018; Opar, et al., 2015; Timmins, Bourne, et al., 2016), therefore it would be proposed that the most important feature would be to understand how to train the hamstrings effectively to lead to positive adaptations that could aid in reducing HSI occurrence within sports.

There is evidence however, that high-speed running is not the only running mechanism within sports, with the potential for acceleration phases within running and running in a flexed hip-position also being a moment where HSIs could occur (Mendiguchia et al., 2014; Morin et al., 2015; Verrall, Slavotinek, & Barnes, 2005). The hamstrings have been proposed to play an important role in the sprint acceleration phase (Edouard, Samozino, Slotala, Mendiguchia, & Morin, 2016; Morin, et al., 2015; Morin, Samozino, & Edouard, 2011), furthermore, acceleration-based tasks occur more frequently within team-based sports (Cahill, Lamb, Worsfold, Headey, & Murray, 2013; Cunniffe, Proctor, Baker, & Davies, 2009; Gabbett and Gahan, 2015; Jones, West, Crewther, Cook, & Kilduff, 2015; Rampinini et al., 2015; Wisbey, Montgomery, Pyne, & Rattray, 2010). Running in a flexed hip-position, which may occur in sports such as rugby or cricket, results in the hamstrings lengthening across the hip joint, which if performed during the terminal swing phase of running would exaggerate further the strain placed upon the hamstrings (Verrall, et al., 2005). As lumbo-pelvic control is another potential risk factor for HSI (Kenneally-Dabrowski, Brown, Warmenhoven, et al., 2019; Opar, et al., 2012; Shield and Bourne, 2018), tasks which involve some lateral flexion of the trunk, such as curved running (Filter et al., 2020), could also heighten the risk of a HSI event due to lengthening of the hamstrings on one side the body. Thus, it could be proposed that the actual mechanism for an injury occurrence could be the lengthening action and the strain placed upon the hamstrings, typically during the late swing phase. This highlights a key consideration regarding training, specifically looking to sprint over a range of distances using both acceleration and high velocity running tasks to prepare athletes (5-, 10-, 20-, 40- and 60 m sprints, 5-20-, 10-10-, 10-15- and 10-20 m flying sprints, 10-10-10-, 20-20-20 m sprint float sprints, 10- and 20 m curved runs).

2.3 Occurrence of Hamstring strain injuries

It should come as no surprise that sports that are reliant on an athlete's top-end speed such as track sprinting, possess the greatest rates of HSIs across any sport (Cross, Gurka, Saliba, Conaway, & Hertel, 2013). This finding was expected due to the proposed mechanism of HSIs, with the most hazardous portion being suggested to be the terminal swing phase of the sprint running gait (Heiderscheit, et al., 2005). Whereas, short acceleration and constant velocity dominant sports, such as racket-based sports and long-distance running, do not place the hamstring MTUs under a similar magnitude of stretch, potentially explaining the lower reported rate of HSIs within these sport (Dalton, Kerr, & Dompier, 2015). Team sport movement demands however are highly variable, with elements of linear and multi-directional top speed, acceleration and deceleration, with individual demands placed upon certain positions (Austin, Gabbett, & Jenkins, 2011a, 2011b; Cahill, et al., 2013; Cummins, Orr, O'Connor, & West, 2013; Cunniffe, et al., 2009; Gabbett and Gahan, 2015; Jones, et al., 2015; Kempton, Sirotic, & Coutts, 2014; Malone, et al., 2018; McLellan and Lovell, 2013; Ruddy, Pollard, et al., 2018; Suarez-Arrones et al., 2014; Wisbey, et al., 2010). This variable nature underpins the high rate of occurrence seen within team-based sports, specifically, with large pitch sports (e.g., football, AFL and rugby) in comparison to smaller court-based team sports such as basketball (Longo et al., 2012). Interestingly, non-running sports, including wrestling, gymnastics, swimming and diving, have the lowest reported incidence of HSI, although HSIs still occurred within these sports. One potential explanation for their occurrences is that these sports require athletes to go through large range of motion (ROM) around the hip and knee joints, when performing gymnastic based tasks or in shots and takedowns from opponents. These tasks could result in micro-scopic muscle damage if athletes have insufficient levels of strength when attempting to apply force within the longest muscle lengths - demonstrating that prevention practices are still important even for non-running sports.

To date, HSI incidence rates (Appendix one) across all sports have been gradually increasing, despite the increasing interest of hamstring related research. One potential explanation could be the fact that many practitioners still avoid performing HSI reducing practices (Bahr, et al., 2015; McCall, et al., 2016; Shield and Bourne, 2018). Furthermore, it could be related to the increased playing intensity and demand of sport, specifically within some leagues and competitions (e.g., European soccer). Where many teams are required to play multiple games within a week, with exceptionally intense periods, including Christmas fixtures in English soccer resulting in increased injury occurrence (Read, Oliver, De Ste Croix, Myer, & Lloyd, 2018; Van Crombrugge, Duvivier, Van Crombrugge, Bellemans, & Peers, 2019; Wollin, Thorborg, & Pizzari, 2017; Wollin, et al., 2018). Additionally, the assessment of injuries over time has also improved, with greater accuracy in the identification of specific injuries, with technologies such as ultrasound (US) and magnetic resonance imaging (MRI), more readily available (Abe, Loenneke, & Thiebaud, 2016; Ekstrand, Lee, & Healy, 2016; Mendiguchia et al., 2013; Wangensteen, et al., 2016).

Although there is some caution that needs to be taken with injury surveillance studies, typically, injury rates are calculated over a specific duration, including 1000 hours exposure (training and match), 10,000 exposures, per season or total exposure. Although comparisons can be made between these methods, they are not identical as 1000 hours could be one season or half a season, depending on the player, season, league or sport. Furthermore, Dalton, et al. (2015) used 10,000 exposures within the rate calculations, with no explanation as to what constitutes an exposure, e.g. could a 15-minute recovery session count as one exposure, which would not equate to a 45 min repeated-sprint workout. A further calculation utilised by Edouard, Branco, & Alonso (2016) used total number of athletes as a measure (rate per 1000 athletes), this difference in calculation could explain the extremely high rates observed for top-end speed running in comparison to other calculations. With regards to future investigations into injury surveillance within sport, reporting should aim for not only a measure of exposure but should also consider an element of intensity, with options such as subjective (player led), objective (coach led), intrinsic loading (heart rate or movement intensities) or extrinsic (rating of perceived exertion). This would enable a more thorough understanding of the loading conditions under which injuries were occurring.

2.4 Risk factors for hamstring strain injury

Several authors have proposed a number of potential risk factors for HSIs, across a variety of team-based sports (Bourne, et al., 2015; Colby et al., 2018; Dauty, Menu, & Fouasson-Chailloux, 2018; Duhig, 2017; Duhig et al., 2016; Fousekis, et al., 2011; Gabbe, Bennell, & Finch, 2006; Green, Bourne, van Dyk, & Pizzari, 2020; Huygaerts et al., 2020; Malone, et al., 2018; Opar, et al., 2015; Stares et al., 2018; Verrall, Slavotinek, Barnes, Fon, & Spriggins, 2001). It has been postulated that risk factors come in multiple forms including intrinsic non-modifiable, intrinsic modifiable and extrinsic modifiable. Examples of intrinsic, non-modifiable risk factors are anatomical factors, age, ethnicity and previous injury (Green, et al., 2020; Opar, et al., 2012), i.e., where practitioners cannot influence or adapt through training and thus, they are ever-present. In contrast the intrinsic, modifiable risk factors are those which can be influenced by training, such as architectural properties, strength (including muscle action force production, motor control and muscle fatigue) and flexibility or mobility. Extrinsic factors include higher levels of match play and competition, ultimately the demands of the sport placed upon the athlete, which can also potentially influence HSI. For the purpose of this thesis, extrinsic factors are not explored, as they are multi-variate often dependent on the opposition, however, it would be presumed that if an athlete was optimally prepared intrinsically (i.e., muscle architecture and strength), then any extrinsic risk would be minimised.

2.4.1 Intrinsic non-modifiable

2.4.1.1 Anatomical factors

The anatomy of the hamstrings potentially predisposes the muscle group to an increased risk of strain injury. The biarticular nature of the hamstrings will result in significant lengthening, when flexion occurs at the hip, with concurrent extension of the knee (Opar, Williams and Shield, 2012), as is seen during the swing phase of running and place kicking (Thelen *et al.*, 2005; Chumanov, Heiderscheit and Thelen, 2011). These lengthening demands are thought to increase the risk of HSI because the lengthening may exceed the mechanical limits of the muscle (Thelen *et al.*, 2005; Chumanov, Heiderscheit and Thelen, 2011).

Further anatomical features of the hamstring, specifically the two heads of the bicep femoris have also been suggested to increase HSI risk (Smet and Best, 2000; Opar, Williams and Shield, 2012). The two heads of the bicep femoris, BF_{LH} and the bicep femoris short head (BF_{SH}), are innervated by two different nerve branches. This dual innervation has the potential for uncoordinated contractions of the BF_{LH} and BF_{SH}, thus increasing HSI risk via poor intra-muscular coordination (Heiser *et al.*, 1984). Recent observations have recently provided some evidence regarding activation coordination and the preferential activation of specific muscles heightening the risk of HSI (Hug et al., 2018; Schuermans, Van Tiggelen, Danneels, & Witvrouw, 2014), the authors suggested that possessing a disassociation and some variability in hamstring muscle recruitment with respect to metabolic activation and a focus on the quality and quantity of hamstring muscle recruitment are critical in HSI risk (Schuermans, et al., 2014; Schuermans, Van Tiggelen, & Witvrouw, 2017). The ST also has a specific anatomy that it is designed for high velocity actions with long fascicles primarily made up of type 2 (fast twitch) fibres and a long distal free tendon possessing a limited amount of slack, these features may dispose the ST to a greater risk of a tendon rupture (Koulouris and Connell, 2003; Beltran *et al.*, 2012).

2.4.1.2 Age

Increasing age has been identified as another risk factor which can be associated with HSI (Gabbe, Bennell, Finch, Wajswelner, & Orchard, 2006; Green, et al., 2020; Henderson, Barnes, & Portas, 2010; Raya-González, De Ste Croix, Read, & Castillo, 2020; Roe et al., 2020; Verrall, et al., 2001; Woods, et al., 2004). The ages of 23 and 24 years have been identified as the point at which HSI risk increases with age across team sports including: AFL, soccer and Gaelic football (Gabbe, Bennell, Finch, et al., 2006; Green, et al., 2020; Henderson, et al., 2010; Roe, et al., 2020; Verrall, et al., 2001; Woods, et al., 2004). Further increases in HSI risk by as much as 1.3 and 1.8-fold, have been reported with increased age in AFL and soccer players respectively (Henderson, et al., 2010; Verrall, et al., 2001). Although training experience or age has also been found to have a protective effect, with the potential for an inverted U relationship between age and HSI risk, with players who have 4-8 years' experience (at the peak of their career) having protective adaptations potentially caused by the demands of training and match play (Duhig, et al., 2016). Interestingly, Roe, et al. (2020) found that within Gaelic footballers, players who were >30 years old had the lowest risk of HSI. Although this finding could be explained by the small proportion of overall sample that was comprised of players >30 years old (5.4% of the total sample), and if a more representative sample was to be used it would be expected to fall in line with previous literature (i.e., greatest risk of injury).

Age-related changes that have suggested to influence increased HSI risk, include; increased body weight (Gabbe, Bennell and Finch, 2006), reduced hip flexor flexibility (Gabbe, Bennell and Finch, 2006; Henderson, Barnes and Portas, 2010), decreased muscle mass (Gabbe *et al.*, 2006) and decreased strength (Gabbe *et al.*, 2006). Within a review by Opar, Williams and Shield (2012), it was noted that the literature supporting these theories included non-elite, non-athletic participants, with greater age ranges than those seen within elite sport (Kirkendall and Garrett, 1998; Doherty, 2001). They followed on by hypothesizing that it would be unlikely that athletes aged 24-30 years are weaker or have less muscle mass than their 18- to 20 year old counter parts (Opar, Williams and Shield, 2012). Notably, studies that have reported age as a risk factor have established this from regression analysis of data and not by looking independently of further variables which may contribute to HSI, including previous HSI event (Opar, Williams and Shield, 2012). This indicates that further research is required to provide a better explanation as to why athletes older than 24 years old, in team based sports are at a significantly greater risk of HSI (Opar, Williams and Shield, 2012), as well as identifying if this trend is replicated across sporting disciplines, including sprint sports.

2.4.1.3 Previous injury

Large detrimental effects have been observed within previously strained hamstrings, with reduced strength, architecture and activation, as well as impacting upon dynamic tasks such as kicking and running (Charlton et al., 2018a; Green, et al., 2020; Higashihara et al., 2019; Kenneally-Dabrowski, Brown, Warmenhoven, et al., 2019; Lord, Ma'ayah, & Blazevich, 2018; Nagano, Higashihara, & Edama, 2015; Silder, Reeder, & Thelen, 2010; Silder, Thelen, & Heiderscheit, 2010; Timmins, Bourne, et al., 2016). Maniar et al. (2016) published an excellent systematic review and meta-analysis highlighting that a previous HSI resulted in clear deficits when athletes returned to play in slow concentric isokinetic peak torque (60 deg/s) and eccentric strength assessed during the NHE. Furthermore, it is unsurprising, reciprocal muscle asymmetries (e.g. hamstring:quadricep ratio), were found to be significantly impacted by a previous HSI across isokinetic velocities (Maniar, et al., 2016). The authors also highlighted that measures of isometric strength and flexibility (passive straight leg raise), were returned to baseline within 20 – 50 days, with a clear reduction in isometric strength observed at only a < 3 days post-HSI (Maniar, et al., 2016), potentially as an effect of pain-driven inhibition. Despite these differences identified within the systematic review and meta-analysis, the deficits are not consistent across the literature, as elite “world class” British sprinters, who had a history of previous HSI, presented no difference in eccentric strength assessed during the NHE. Furthermore, they actually presented stronger than those with no history of HSI (Giakoumis et al., 2020), these contrasting findings could be explained by the elite sprinters having a greater strength training status in addition to greater resources provided for strength and conditioning provision. Indicating, that training to reach the upper echelons of sport, including sprinting could have the potential to mitigate the effect of HSIs. In addition, no differences were also observed across other methods of hamstring strength assessments, including; higher velocity concentric and eccentric isokinetic tasks (Maniar, et al., 2016).

Although there is limited evidence surrounding the structural changes associated with a previous HSI, BF_{LH} architecture has been found to be reduced in those with a previous HSI (Timmins et al., 2017). Timmins et al, (2017) found that within elite Australian soccer players, those with a previous HSI possessed shorter BF_{LH} fascicle lengths (FL), in comparison to the contralateral non-injured limb and to a control group consisting of non-injured elite soccer players at all time points across a season, with moderate to large effects (Cohen’s d = 0.76 –

1.15). However, as highlighted this study has only been performed by one research group in a single team sport and requires replication to reinforce this relationship. Further exploration of the effect on a reduced BF_{LH} FL contributing to an elevated risk of a HSI event is detailed within this literature review, including various methodological issues. However briefly, the FL of the muscle can influence the muscles force-velocity and force-length relationships (Timmins, Shield, Williams, Lorenzen, & Opar, 2016), which can hamper its ability to control the rapid lengthening of the hamstrings that occurs during the terminal swing phase of running across the muscle's descending limb of the force-length curve (Chumanov, et al., 2011; Thelen, Chumanov, Best, et al., 2005; Timmins, Shield, Williams, Lorenzen, et al., 2016).

Despite these intrinsic, non-modifiable risk factors being important considerations for practitioners, the intrinsic modifiable risk factors can ultimately reduce the risk of sustaining a future HSI by meaningful positive changes (Giakoumis, et al., 2020; Pizzari, et al., 2020). Therefore, they should be the primary focus for practitioners and will also be the focus of this thesis. Further to this, as mentioned, optimal preparation in the form of appropriate architectural properties and sufficient levels of strength, would ultimately reduce the influence of the extrinsic modifiable risk factors. For instance, optimal preparation of an athlete would reduce the risk of HSI events particularly during youth to senior transitions, where youth athletes, who are typically weaker step-up to senior higher levels of match play (Franchi et al., 2019; Roe et al., 2018), or even transfers across leagues where match demands are variable (Aughey, 2011; Cahill, et al., 2013; Kempton, et al., 2014; Mendez-villanueva, Buchheit, Simpson, & Bourdon, 2012; Owen, Venter, du Toit, & Kraak, 2015; Suarez-Arrones, et al., 2014; Wisbey, et al., 2010).

2.4.1.4 Gender

To date the majority of injury epidemiology studies have been fairly one sided, with a focus on male athletes, particularly team sport athletes such as football, rugby and AFL. Although with the growing number of professional female athletes and leagues and greater financial support permitting an improvement in available resources, it would be suspected in the future a more balanced reporting of injuries within research could be achieved. The imbalance in reporting, has led to suggestions that male athletes are more susceptible to a HSI (Cross, et al., 2013; Dalton, et al., 2015). The American collegiate sport system provides a high level of availability of resources (i.e. strength and conditioning, physiotherapy and athletic training/rehabilitation) across a range of sports available for both males and females. Dalton, et al. (2015) observed the HSI occurrence across six National Collegiate Athletic Association sports, with both male and female participants (indoor track, outdoor track, soccer, lacrosse, ice hockey and basketball). The authors identified that males sustained HSIs more frequently than females in indoor track, outdoor track, soccer and lacrosse, with higher HSI occurrences within ice hockey and basketball for female athletes (Dalton, et al., 2015). Similar findings were also reported by Cross, et al. (2013) and Edouard, Branco, et al. (2016), within soccer and track sprinting respectively, with a greater HSI incidence within male athletes.

Despite a higher occurrence in some sports within males, females still sustained HSIs (Cross, et al., 2013; Dalton, et al., 2015; Edouard, Branco, et al., 2016), therefore, HSI prevention practices are still an important consideration for training female athletes. Additionally, the traditional high HSI injury occurrence sports in male athletes (soccer and track), females still sustain HSIs more frequently than other lower incidences sports such as lacrosse, ice hockey and basketball. Moreover, with the increased resources being provided to female athletes including improved strength and conditioning provision, it would be presumed the overall intensity of the training and matches would also increase especially movement intensities such as achieving higher running which further increase the rate of HSIs. It should also be noted that the hamstrings have function, specifically in preventing severe knee injuries. This in increasingly important for female athletes who are more likely to sustain an anterior cruciate ligament (ACL) rupture than male counterparts (Boden, Sheehan, Torg, & Hewett, 2010; Lloyd, 2002; B Yu and Garrett, 2007). The hamstrings are proposed to be aiding in ACL risk reduction, via reducing anterior tibial translation, traditionally thought to be via isometric-eccentric muscle action during high risk tasks such as change of direction or jump landings (Withrow, Huston, Wojtys, & Ashton-Miller, 2008), highlighting potential similarities in prevention practices for both HSIs and ACL injuries.

2.4.2 Intrinsic modifiable

2.4.2.1 Bicep femoris fascicle length

The architecture of the active components of a muscle, i.e. FL, muscle thickness (MT), pennation angle (PA), and cross-sectional area (CSA) dictates a muscles force generating potential – this includes both a muscles force-velocity and force-length relationships (Timmins, Shield, Williams, Lorenzen, et al., 2016). A single muscle fascicle is a chain of in-series sarcomeres; therefore, a longer fascicle is comprised of an increased number of sarcomeres which permits a greater shortening velocity (Timmins, Shield, Williams, Lorenzen, et al., 2016). In addition, the increased number of sarcomeres would permit an increased working range, specifically increasing the active and passive force generating capacity on the descending limb of the force-length relationship, thus reducing the potential for muscle damage (Timmins, Shield, Williams, Lorenzen, et al., 2016). This is of particular importance for the descending limb of the force-length curve of the hamstrings, which occurs within the terminal swing phase of running, with the potential for micro- and macroscopic muscle damage, which are proposed mechanisms for HSIs (Morgan and Proske, 2004). Despite sarcomerogenesis, which is the hypothesised process of the addition of sarcomeres within-series, having a profound effect on improving muscle function and the potential for reducing the risk of injury, it has been documented that across a single muscle and muscle fascicle sarcomeres are typically non-uniform under resting and active conditions, and what is currently unknown is how training influences sarcomerogenesis and the non-uniformity of sarcomeres (Moo, Leonard, & Herzog, 2017) (Figure 2-1) .

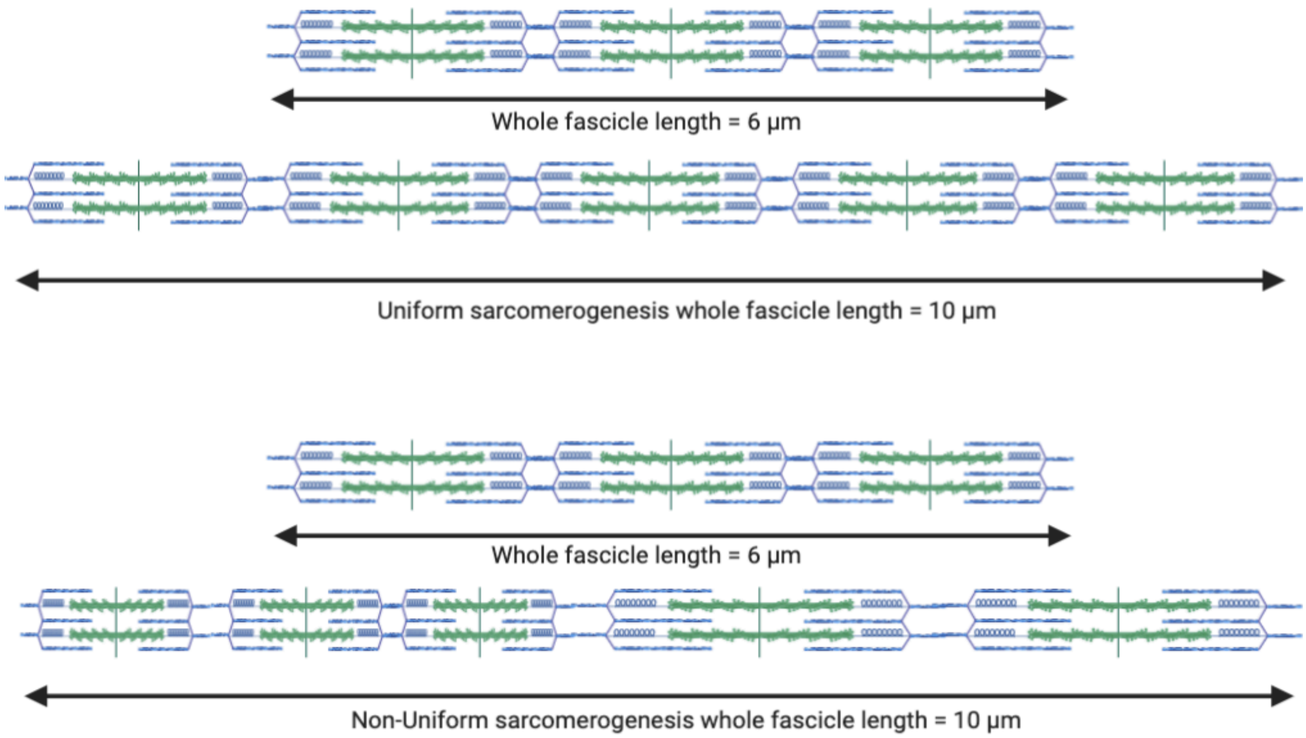


Figure 2-1 A single muscle fascicle with uniform and non-uniform sarmerogenesis.

The influence that architecture has on the functioning characteristics of a muscle, has thus established FL as a modifiable risk factor for HSI – specifically within BF_{LH}. FL has been identified as an important risk factor of HSI in soccer players (Timmins *et al.*, 2016), particularly when associated with low levels of eccentric knee flexor strength (Timmins *et al.*, 2016; Bourne *et al.*, 2017) although there is limited association between the two measures. One hypothesis as to why shorter fascicle are predisposed to a greater risk of HSI, is due to damage to the sarcomere via 'popping' which occurs while being lengthened (Morgan, 1990; Morgan and Proske, 2004), for instance during the terminal swing phase of running (Thelen *et al.*, 2005; Higashihara *et al.*, 2010; Chumanov, Heiderscheit and Thelen, 2011; Nagano *et al.*, 2014). Due to the reduced number of within series sarcomeres of shorter fascicles, there is a greater potential for increased eccentric muscle damage i.e., "popping", as there are a fewer number of sarcomeres and cross bridges to "pop". Therefore, shorter fascicles would endure greater muscle damage or microscopic damage leading to a single traumatic event during high force, lengthening actions (Opar, Williams and Shield, 2012; Bourne *et al.*, 2017).

At rest, a BF_{LH} FL of <10.56 cm, a FL relative to measured MT of the BF_{LH} of <0.254, and BF_{LH} FL at 25% of a maximal contraction of <9.61 cm were all considered significant risk factors for future HSI in elite soccer players (Timmins *et al.*, 2016), with an increased risk (risk ratio (95% confidence intervals) of future HSI occurrence of 4.1 (1.9-8.7), 3.7 (1.9-7.3) and 3.2 (1.2-7.9), respectively (Timmins *et al.*, 2016). The values presented by Timmins et al.(2016), should however be interpreted with caution and cannot be applied categorically across all sports. Firstly, these values are population specific, this not only includes for team sport athletes – with the potential for increased values for sprint athletes, who naturally possess greater FLs (Abe, Kumagai, & Brechue, 2000; Kumagai et al., 2000). Furthermore, as the values presented by Timmins et al.(2016) are only specific for Australian elite soccer, which in comparison to elite levels (e.g. European leagues) is of a reduced match intensity (Cummins, et al., 2013; Gabbett, Stein, Kemp, & Lorenzen, 2013; Malone, et al., 2018; Malone et al., 2017; Mendez-villanueva, et al., 2012; Rampinini, et al., 2015; Rampinini et al., 2007; Stølen, Chamari, Castagna, & Wisløff, 2005; Wisbey, et al., 2010). This could indicate that in order to mitigate HSI occurrence in European soccer players, greater FLs than those provided could be required. Secondly, there are a number of methodological inaccuracies within the methods of estimating BF_{LH} FL (Franchi, Fitze, Raiteri, Hahn, & Spörri, 2019; Franchi et al., 2018; Freitas, Marmeleira, Valamatos, Blazevich, & Mil-Homens, 2018; Pimenta, Blazevich, & Freitas, 2018); the methods for assessing resting BF_{LH} will be explored further in more detail later within this literature review.

2.4.2.2 Strength

Stronger muscles have been proposed to provide greater protection against muscle strains (Garrett *et al.*, 1987; Suchomel, Nimphius and Stone, 2016), which would facilitate in making strength a modifiable risk factor in HSI (Opar, Williams and Shield, 2012). However, as with injuries in general, the overall effect of strength is extremely multifaceted with several proposed sub-factors related to strength (Figure 2-2). As with any risk factor, there is always a varied level of evidence, this is especially true with strength with a number of sub-factors. Moreover, recent advancements in technology have resulted in a "boom" of literature providing novel evidence of potential strength related risk factors for HSI.

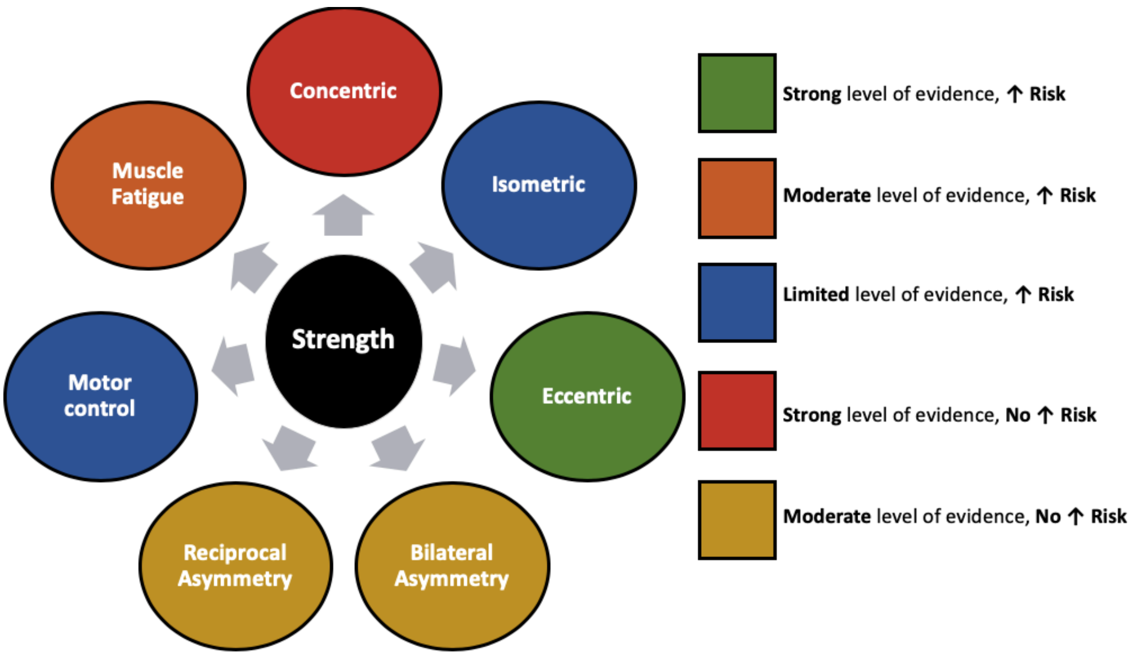


Figure 2-2 Sub-risk factors associated with hamstring strength, with levels of evidence and relationship to injury risk identified. Adapted from (Pizzari et al., 2020)

2.4.2.2.1 Muscle action type

Within research and practice, the assessment of hamstring strength has been performed across all three muscle actions; using various pieces of equipment to facilitate this (isokinetic dynamometer, strain gauges and force platforms). From the early conception of athlete profiling, typical practice incorporated concentric strength assessments (Croisier, Ganteaume, Binet, Genty, & Ferret, 2008; Green, et al., 2018; Higbie, Cureton, Warren iii, & Prior, 1996; Linnamo, Moritani, Nicol, & Komi, 2003; Sugiura, Saito, Sakuraba, Sakuma, & Suzuki, 2008). However, as time and knowledge around the various muscle actions the focus has shifted onto using eccentric strength assessments using gold standard laboratory methods (isokinetic), along with more novel field-based measures of eccentric strength assessment (Chalker, et al., 2016; Franchi, Ellenberger, et al., 2019; Green, et al., 2018; Lee, Cai, Yung, & Chan, 2018; Opar, Piatkowski, et al., 2013; Opar, Williams, Timmins, Dear, & Shield, 2013b; Opar, et al., 2015; Roe, Malone, et al., 2018; Timmins, Bourne, et al., 2016; van Dyk, Witvrouw, & Bahr, 2018; Wiesinger, Gressenbauer, Kösters, Scharinger, & Müller, 2019). Additionally, with a recent increase in the availability and advancement of force plate technology, isometric assessments have become more common practice within elite sport (Charlton, et al., 2018a; Charlton et al., 2018b; McCall et al., 2015; Wollin, et al., 2016; Wollin, et al., 2017, 2018), with sport science and strength and conditioning practitioners citing an increased speed of assessment, along with a reduction in post-testing fatigue.

2.4.2.2.2 Concentric

Despite stronger muscles providing greater protection to injury (Suchomel, Nimphius, & Stone, 2016), across all concentric isokinetic velocities (60-300 deg/s), there is a strong level of evidence demonstrating that both absolute and relative strength qualities have no association to HSI occurrence (Green, et al., 2020; Green, et al., 2018). Further supporting evidence demonstrated that concentric hamstring strength at isokinetic velocities of 60- and 180 deg/s had no significant relationship with HSI occurrence in professional soccer players (Fousekis, et al., 2011). One potential explanation as to why concentric assessments may not hold predictive validity, is that concentric knee flexor strength appears to be unaffected by HSI occurrence, with similar isokinetic strength profiles between injured and non-injured limbs (Bennell et al., 1998; Sugiura, et al., 2008). Although the time delay between injury and assessment was not documented by the authors which could have influenced this finding. However, Van Dyk and colleagues (2016) made similar observations within professional soccer players where concentric strength values were similar between pre- and post-injury, where the post-injury assessment took place within 7-days of a HSI incidence. Isokinetic values were similar at return to sport, with a 4.1% difference from pre-injury levels at 60 deg/s, which was significant (p = 0.03); although there was no difference observed at faster isokinetic velocities (300 deg/s) (Van Dyk, et al., 2016). However, it should be noted that the range of percentage differences between pre-injury and return to sport isokinetic strength was extensive, with values between 68.3-152.5% for 60 deg/s and between 60.3-127.8% for 300 deg/s (Van Dyk, et al., 2016). These results demonstrate that there is a large degree of individual variance for concentric strength across time points of testing (e.g. pre-, post-injury and at return to sport), with the potential for a large strength deficit at return to sport which could be increasing the risk of future HSI, explaining why a previous HSI is a primary risk factor in the re-occurrence of an HSI event (Opar, et al., 2012). Despite concentric assessments lacking predictive validity for HSI incidence (Green, et al., 2020; Green, et al., 2018).

2.4.2.2.3 Eccentric

Lengthening actions were proposed to be more specific to the mechanisms of HSI, therefore eccentric assessments were proposed to possess a greater predictive effect to future HSI, which is one reason for the shift to eccentric assessments to align with the muscle action experienced during the terminal swing phase. A recent meta-analysis examining the predictive validity of isokinetic strength assessments to detect the risk of future HSI, found that absolute and relative eccentric isokinetic assessments at 60 deg/s had a predictive effect, albeit small in magnitude (Green, et al., 2018). A recent "recalibration" of HSI risk (Green, et al., 2020), performed by the same authors of the previous meta-analysis (Green, et al., 2018), demonstrated that eccentric hamstring strength remained associated with the risk of HSI. A similar observation was found by Fousekis, et al. (2011), whereby the greatest odds ratio of future HSI was found with eccentric hamstring strength asymmetries – i.e., one leg being weaker than the other, although the specific isokinetic velocity was not identified with both 60- and 180 deg/s being assessed. However, not all studies have supported the consensus that eccentric isokinetic values have a significant relationship with future HSI incidence (Bennell, et al., 1998; Green, et al., 2018; Lee, Mok, Chan, Yung, & Chan, 2018; Van Dyk, et al., 2016; van Dyk, Witvrouw, et al., 2018; Yeung, Suen, & Yeung, 2009). Early research by Bennell et al. (1998) found that isokinetic assessments could not discriminate between those with and without injury, therefore, they are unable to predict the likelihood of future HSI within team sport athletes.

Despite inconclusive results regarding eccentric isokinetic assessments providing a strong predictive validity for detecting risk of future HSI. Isokinetic assessment remains the gold-standard lab-based methods of assessing single joint strength (Aagaard, Simonsen, Magnusson, Larsson, & Dyhre-Poulsen, 1998; Bennell, et al., 1998; Croisier, et al., 2008; van Dyk et al., 2018; van Dyk, Witvrouw, et al., 2018; Zakas, 2006; Zvijac, Toriscelli, Merrick, & Kiebzak, 2013). However, as lab-based measures are difficult to perform with large groups of athletes in a time efficient manner, a field-based device for eccentric hamstring assessment was designed (Opar, Piatkowski, et al., 2013). The "Nordbord" utilises strain gauges situated at the ankle, which assesses the vertical force produced at the ankle when the NHE or alternative assessments are performed (Opar, Piatkowski, et al., 2013). The Nordbord has been found to be a reliable method of assessing eccentric hamstring strength (Opar, Piatkowski, et al., 2013; Opar, Williams, Timmins, Dear, & Shield, 2013a), where the authors identified that within elite AFL players, low eccentric strength was a risk factor for future HSI (Opar, et al., 2015).

Typical scores achieved upon the Nordbord range from ~200 N to ~700 N for youth up to senior elite team sport athletes (Bourne, et al., 2015; Buchheit, Cholley, Nagel, & Poulos, 2017; Chalker, et al., 2016; Franchi, Ellenberger, et al., 2019; Opar, et al., 2015; Roe, et al., 2020; Roe, Malone, et al., 2018; Timmins, Bourne, et al., 2016; van Dyk, Witvrouw, et al., 2018), current research suggests ~350 N being a functional minimum whereby the risk of future HSI is reduced for elite soccer athletes (Bourne, et al., 2018; Opar, et al., 2015; Timmins, Bourne, et al., 2016). However, a recent meta-analysis came to a contrasting conclusion, where eccentric hamstring strength as assessed on the Nordbord was not related to HSI incidence (Green, et al., 2020). This highlights that evidence might not be as clear cut as initially thought, with the authors suggesting that a single test occasion could be inadequate and ongoing monitoring would be more effective (Green, et al., 2020). Despite some inconsistencies, both the Nordbord and isokinetic methods can be utilised to assess eccentric hamstring strength, with some potential to discriminate potential risk of HSI.

Both in research and practice, common practice for the "Nordbord" is to utilise bodyweight alone in hamstring assessments, despite bodyweight being capable of explaining up to 24% of the overall variance within Nordbord assessments (Buchheit, et al., 2017). However, the utilisation of a progressive one repetition maximum assessment known as the Nordic eccentric strength test (NeST) is becoming more frequent (Duhig et al., 2019). The NeST involves the performance of single repetitions of the NHE upon the Nordbord with the progressive addition of external loads (+5 kg), held at the xiphoid process until the force produced did not increase. Anecdotally, the addition of load during the NeST will improve the observed forces for stronger individuals, whereas in contrast weaker individuals the forces would be expected to reduce – as they would be unable to tolerate the increased demand. Despite providing a beneficial effect for stronger individuals, it is currently unclear if the addition of during testing can differentiate between those at an increased risk of injury. Furthermore, as it alters individuals force production on an initial strength basis, it prevents comparison between individuals (Bourne, et al., 2015; Bourne, et al., 2018; Duhig, et al., 2019). Therefore, research and practice should continue to assess the NHE using bodyweight alone when screening an individual's eccentric hamstring strength. But when it comes to training the addition of load should be considered integral to achieve the greater intensities and as with classical lifting maximums, it might be prudent to assess individuals using the NeST to determine appropriate training loads.

An alternative method of assessing eccentric hamstring strength of athletes, is to perform video analysis of the NHE, which is not unlike the Nordbord, but this method presumes that the break point angle achieved is a proxy for eccentric hamstring strength (Lee, Cai, et al., 2018; Lee, Li, Yung, & Chan, 2017). This presumption is supported within the literature with large – very large, significant associations between break-point angle and eccentric peak hamstring torque at both 30- and 60 deg/s , in addition there was a very large and significant association between break-point angle and relative eccentric peak torque (30 deg/s) (Lee, Cai, et al., 2018; Lee, et al., 2017; Sconce, Jones, Turner, Comfort, & Graham-Smith, 2015). Significant relationships were also found with concentric hamstring peak torque at 60 deg/s, albeit they were weaker in magnitude (moderate – large) than eccentric actions (Lee, Cai, et al., 2018; Lee, et al., 2017), which is understandable the NHE is an eccentric based exercise. The highlighted associations indeed appear to corroborate the fact that the break point angle achieved within the NHE is related to both eccentric and concentric hamstring strength. Therefore, it could be used as an alternative method of assessing hamstring strength, while able to discriminate between heavier and taller athletes - although the limited associations with concentric strength capabilities in contrast to eccentric measures may reduce its effectiveness if looking to observe concentric strength qualities. Although further research is required to determine if a measure of break point angle achieved within the NHE, is sensitive enough to discriminate the potential risk of HSIs within athletes.

2.4.2.2.3.1 Comparison between assessment methods

Despite both isokinetic and NHE eccentric strength assessments claiming to assess identical muscular qualities, i.e., eccentric knee flexion, research currently suggests that they may not be assessing identical qualities. Recently, a Nordbord type device has been shown to have a poor relationship (r = 0.51 – 0.58) with eccentric isokinetic assessments, when the device had both controlled and non-controlled movement velocities (Wiesinger, et al., 2019). The authors suggested that the Nordbord eccentric measures are made in a ROM that does not include the actual angle of peak torque (Wiesinger, et al., 2019). As the NHE does not reach an angle of peak torque or moment, the assessor does not know what the maximum is that could be attained, unlike the dynamometer. Furthermore, the force achieved during the NHE depends on the gravitational moment achieved prior to the breakpoint, highlighting it is entirely different to isokinetics (van Dyk, Witvrouw, et al., 2018; Wiesinger, et al., 2019).

The authors go on to suggest that equipment which enables greater ROM may be necessary moving forward to increase the validity of Nordbord type assessments (Wiesinger, et al., 2019), such as the recently recommended device by Giacomo et al, (2018). This is potentially due to changes in hip angle including MTU and FL curve, influencing the force output. Despite these findings and the suggestions by Weisinger and colleagues (2019) the efficacy of using the angle of peak torque in HSI risk measurement and rehabilitation has yet to be fully established (Timmins, Shield, Williams, & Opar, 2016). Therefore, utilising the Nordbord device to assess force generating characteristics can be a useful and practical assessment, with research supporting its use in HSI risk measurement (Bourne, et al., 2015; Bourne, et al., 2018; Opar, Piatkowski, et al., 2013; Opar, Williams, et al., 2013a; Opar, et al., 2015; Timmins, Bourne, et al., 2016). Yet, comparisons between methods (isokinetics vs. Nordbord) should, at the very least, be made with caution or not compared at all, rather looked upon as alternative methods of assessment both telling varying force-length characteristics of the hamstrings, while both offering a potential HSI risk assessment (Wiesinger, et al., 2019). Despite the shortcomings with the assessment itself, there are key logistical differences which may explain their use, specifically the time and efficiency of assessing large numbers of athletes and the availability of an isokinetic dynamometer.

2.4.2.2.4 Isometric

Despite limited evidence into its relationship with HSI occurrence, it would be suggested that improvements in isometric strength would offer a protective benefit to injury occurrence (Charlton, et al., 2018a, 2018b; Green, et al., 2020), as stronger muscles would be expected to sustain injury type events less frequently or as Louie Simmons describes in its simplest form "weak things break". However, with the advancement and increased availability and affordability of technologies, including force plates, strain gauges, novel testing devices and hand-held dynamometers (McCall, et al., 2015; Wollin, et al., 2016; Wollin, et al., 2017, 2018). This has resulted in a number of assessment protocols being established within the literature, with a variety of joint angles, muscle lengths and anatomical positions being adopted (McCall, et al., 2015; Wollin, et al., 2016; Wollin, et al., 2017, 2018).

McCall et al. (2015), utilised a force plate at two different knee angles (30- and 90° of knee flexion) to assess isometric hamstring strength, these angles were chosen due to the activation profiles identified within previous literature (McCall, et al., 2015) – although only knee angles were taken into account with respect to overall muscle lengths, despite the hamstrings being a biarticular muscle. High reliability was observed for both knee angles (Intraclass correlation coefficient (ICC) >0.86, coefficient of variation (CV) <6.31%), although the dominant leg at 30° had the largest confidence limits and greatest variability (McCall, et al., 2015). Immediately post-competitive 90 min match, isometric scores for a sub-sample were meaningfully reduced (-29.3 to -50.8 N), to a greater extent than the minimal differences identified (McCall, et al., 2015). Potentially highlighting that isometric assessment could be used to identify potential fatigue, which is a risk factor for injury.

Moderate to high intra- and inter-day reliability has also been shown for externally fixed dynamometers (Charlton, et al., 2018a, 2018b; Wollin, et al., 2016). Furthermore, these devices have been shown to be sensitive enough to identify meaningful decreases in knee flexion isometric strength at varying time points post-competitive team sport matches (Charlton, et al., 2018b), as well as during periods of high match congestion (Wollin, et al., 2018), with large individual variance in force production up to 74 hours post-match (Charlton, et al., 2018b). Although, there were methodological differences between the studies, whereby, knee flexion angle was not standardised within the study by Charlton and colleagues (2018b), in place taking the shank to parallel to the floor, which would be influenced by stature. More recently, an externally fixed dynamometer assessing isometric knee flexion strength, was found to potentially have some predictive ability to identify an athlete's history of HSI (Wollin, et al., 2018). Yet, how useful would this information be for practitioners? Where even in semi-professional sports detailed injury history reports are available, along with simple questioning of an athlete, i.e., "have you had a previous HSI?". Although it has been experienced previously within elite sport where players often have trouble remembering injuries or even which side.

2.4.2.2.5 Strength Asymmetries

Lower limb asymmetries occur as a result of leg length imbalance, injury history and imposed sport demands (Bishop, Read, Chavda, & Turner, 2016). Asymmetries could have an impact on performance, along with increasing the risk and incidence of injury in team sports (Bishop, et al., 2016; Bourne, et al., 2015; Opar, et al., 2012).

2.4.2.2.5.1 Bilateral Asymmetry

When the hamstring from one leg is significantly weaker than the contralateral leg, known as hamstring bilateral asymmetry, it has been suggested to predispose the weaker hamstring to a higher risk of injury (Opar, et al., 2012). One possible explanation for a greater risk of HSI is that bilateral asymmetries may impact on running biomechanics causing different loading to occur across the hamstrings (Lee, Reid, Elliott, & Lloyd, 2009). Several studies have demonstrated that a bilateral strength asymmetry leads to an increased HSI risk (Bennell, et al., 1998; Croisier, 2004; Croisier, et al., 2008; Dauty, et al., 2018; Fousekis, et al., 2011; Freckleton and Pizzari, 2013; Gabbe, Finch, Bennell, & Wajswelner, 2005; Green, et al., 2018; Lee, Mok, et al., 2018; Opar, et al., 2012; Orchard, Kountouris, & Sims, 2017; Orchard, Marsden, Lord, & Garlick, 1997; Verrall, et al., 2001). With asymmetries of ≥ 8% in AFL players (Orchard, et al., 1997), > 10% asymmetry American football and track and field athletes

(Higashihara, et al., 2019; Zvijac, et al., 2013) and > 15% asymmetry in soccer players (Croisier, et al., 2008), increasing the risk of future HSI. However, there is no consensus between the studies with bilateral asymmetry, as Australian rules footballers and sprinters showing no influence of HSI incidence from asymmetry (Bennell, et al., 1998; Yeung, et al., 2009). Nordbord type devices are also capable of assessing bilateral asymmetry (Bourne, et al., 2015; Chalker, et al., 2016; Opar, Piatkowski, et al., 2013; Timmins, Bourne, et al., 2016). Within soccer and rugby union players, a greater bilateral imbalance of eccentric peak force was associated with a greater risk of HSI (Bourne, et al., 2015; Timmins, Bourne, et al., 2016), however the same was not found within Australian Rules football players (Opar, et al., 2015).

A major drawback within the present literature around bilateral asymmetries is that many studies have looked at various sporting populations, with potentially large differences in their imposed competitive sporting and training demands, influencing the observations found within bilateral asymmetries (Bourne, et al., 2015; Opar, et al., 2015; Timmins, Bourne, et al., 2016; Yeung, et al., 2009). Furthermore, as a range of outcome measures have been utilised to determine bilateral asymmetries within athletes, to effectively discriminate between those at a risk of future HSI, researchers should aim to establish normative values for the population under investigation then determine who is outside the norm. However, this is commonly not performed within research potentially due to logistical difficulties as it would require extremely large sample sizes and long observation durations. Moreover, the use of bilateral task in determining asymmetries and the potential of a HSI could also be flawed, as it would not only have a poor relationship to a unilateral task as it is also highly unlikely that a HSI event would involve both limbs simultaneously.

2.4.2.2.5.2 Agonist-Antagonist Asymmetry

A hamstring to quadriceps (H:Q) ratio represents the difference in maximal strength between the two muscle groups, during flexion and extension of the knee i.e., agonist-antagonist asymmetries. A lower H:Q ratio has been suggested to demonstrate that the hamstrings have a poor capacity to resist knee extension during the swing phase of running (Opar, et al., 2012), which may lead to greater angular momentum during the swing phase exceeding the mechanical limits of the hamstring (Higashihara, et al., 2016; Opar, et al., 2012). The conventional method of calculating H:Q ratio is to assess the concentric strength imbalances across the knee (extension and flexion); however, it has been noted a possible flaw with this method is that it is not representative of the eccentric muscle action that the hamstrings could be performing during the terminal swing phase of running (Opar, et al., 2012). Therefore, calculating the H:Q ratio using concentric quadriceps and eccentric hamstring strength measures (functional H:Q ratio) is more relevant to the muscle action performed during the terminal swing phase of running (Aagaard, et al., 1998; Graham-Smith, Jones, Comfort, & Munro, 2013; Opar, et al., 2012).

The difference in methods and sports, has led to a variety of ratios being identified to elevate HSI risk, within American football, a conventional ratio of less than 0.50 (i.e. the hamstrings are 50% weaker than the quadriceps), was found to signify an elevated risk of HSI (Burkett, 1970; Heiser, Weber, Sullivan, Clare, & Jacobs, 1984), while in Australian rules football a conventional ratio of less than 0.61 was determined to put individuals at a substantially greater risk of HSI (Orchard, et al., 1997). Yeung, Suen and Yeung (2009) identified that sprinters who proceeded to suffer a HSI during their season had a conventional H:Q ratio of 0.71, whereas sprinters who remained uninjured had a ratio of 0.96 using the functional method. Croisier and colleagues (2008), used a large scale study (*n*=462) including professional soccer players from multiple leagues (Belgian, Brazilian and French). Identifying that an imbalance of strength with a low H:Q ratio (H:Q conventional <0.45-0.47, H:Q functional, <0.80-0.89) was a risk factor for professional soccer players, and that with restoration of this imbalance could significantly decrease the risk of injury (Croisier, et al., 2008). The sporting demands could be a key factor within H:Q ratio thresholds, from the values presented within the literature, the sports with the greatest high speed running demands require the greatest H:Q ratios (soccer up to sprinting), although there is the potential for within sporting populations and positions, such as American football, to require greater ratios (e.g., wide receivers, running backs may perform longer running bouts (higher running velocities whereas in contrast defensive or offensive line where short accelerative sprints are the primary movement demands). Although not all of the literature using H:Q ratios is in agreement; Bennell et al.(1998) observed contrasting results - where either method (convention or functional) could not discriminate between injured and non-injured limbs in Australian rules football. However, the results of the study by Croisier and colleagues is highly significant in HSI prevention, due to the large, multi-country sample that was utilized (Croisier, et al., 2008), identifying and correcting muscle strength imbalances can reduce the risk of HSI (Croisier, et al., 2008).

Conversely, if an individual has a low H:Q ratio, they therefore possess weak hamstrings – consequently in its simplest form the hamstrings are weak and require strengthening. This is especially important as it is a ratio score and even a perfect (low risk) ratio might mask an individual's inherent weakness if the both the quadriceps and hamstrings individual peak torque results are not observed independently, for instance a high H:Q ratio could indicate weak quadriceps but is masked by the ratio. This highlights one of the key limitations within the H:Q ratio, something that is underexplored or misunderstood is how concentric knee extension could be involved in HSI risk. Specifically, the action where HSI are proposed to occur despite containing knee extension, it is in fact decelerative phase and it is highly unlikely a concentric knee extension action is limiting or producing an effect on the joint kinematics or kinetics (Alt et al., 2020; Edouard, Samozino, et al., 2016; Kenneally-Dabrowski, Brown, Warmenhoven, et al., 2019; Morin et al., 2012; Morin, et al., 2015; Nagahara, Mizutani, Matsuo, Kanehisa, & Fukunaga, 2018; Schuermans, Danneels, Van Tiggelen, Palmans, & Witvrouw, 2017; Schuermans, Tiggelen, Palmans, Danneels, & Witvrouw, 2017). One key practical application when observing a ratio score is to look at the component measures of the ratio, to provide a full picture, in this case the muscle or force generating capabilities crossing the knee joint. It could be hypothesised concentric hip flexion and eccentric knee flexion, potentially as an agonist-antagonist ratio, may have a stronger association with HSI due to the greater involvement in sprint gait, especially in the late swing phase (Guex, Gojanovic, & Millet, 2012; Guex and Millet, 2013; Handsfield et al., 2017; Kakehata, Goto, Iso, & Kanosue, 2020; Mero, Komi, & Gregor, 1992; Nagano, et al., 2014).

2.4.2.3 Neuromuscular function and Motor control

Neuromuscular function is essential for optimal motor output and optimal performance. As HSIs are proposed to commonly occur during the late swing phase of running, an optimal motor output required to counteract the accelerating shank would include an appropriately timed counteracting action (application of force), consisting of a high rate of force or torque development from the hamstrings to decelerate the shank (Chumanov, et al., 2011; Thelen, Chumanov, Best, et al., 2005; Thelen, Chumanov, Hoerth, et al., 2005). Prospectively however, neuromuscular function including rate of torque development and onset of activation, has been shown to have a minimal influence on the risk of HSI incidence (van Dyk, Bahr, et al., 2018). Although the results found by van Dyk, Bahr, et al. (2018) should be interpreted with caution as this study is not without its limitations, with the high potential for fatigue through the testing protocol (15 maximal concentric isokinetic repetitions were performed prior to eccentric assessment), especially as we should consider the eccentric isokinetic assessments with more regard as they have a higher predictive validity and association with HSIs. The potential for fatigue could be an explanation for the high variability observed within the study for rate of torque development (van Dyk, Bahr, et al., 2018), within-sample variability (CV) for eccentric rate of torque development between 65.8 -71% for the injured limb and 47.1-56.5% for uninjured players. Furthermore, rate of torque development has been shown to be an unreliable measure of isokinetic performance, with poor between session reliability and large error within the measurement (Grindstaff et al., 2018), being highly influenced by the angle of peak torque.

Retrospectively, a number of studies have found differences in aspects of motor control (Avrillon, Hug, & Guilhem, 2020; Buhmann, Trajano, Kerr, & Shield, 2020; Schuermans, Danneels, et al., 2017); specifically, these studies highlighted that motor control is negatively affected by a HSI. With proximal control mechanisms potentially aiding in HSI protection (Schuermans, Danneels, et al., 2017), suggesting that motor control and its influence upon lumbopelvic-hip co-ordination, can impact HSI risk (Schuermans, Van Tiggelen, et al., 2017). Moreover, there are also potential deficits in central nervous system processing having an negative influence on motor control and HSIs (Yamada and Mastumoto, 2009). However, it should be highlighted, that the current level of evidence for is the influence of neuromuscular function and motor control as a risk factor is limited and requires further exploration.

2.4.2.4 Muscle fatigue

Unsurprisingly, when a muscle is fatigued the amount of energy required to lead to structural failure, is reduced (Opar, et al., 2012). With a number of HSIs occurring during later stages of match play within team sports and during periods of high match congestion (Brooks, et al., 2005; Cross, et al., 2013; Woods, et al., 2004). Fatigue can manifest itself in both acute stages immediately after a high intensity bout, it can also be formed over longer chronic periods such as periods of high match congestion where inadequate recovery is provided, such as with Selye's general adaptation theory (Haff, 2016). With both acute and chronic fatigue, work induced muscle damage, a reduced central nervous system capacity and potential of an altered circadian rhythm are leading causes of fatigue (Reilly, Drust, & Clarke, 2008; Temesi et al., 2013). It has been proposed that fatigue, from match play and training, can result in altered biomechanics, neuromuscular control and reduced strength capabilities (Lord, Blazevich, Drinkwater, & Ma'ayah, 2018; Lord, Ma'ayah, et al., 2018; Pinniger, Steele, & Groeller, 1999; Roksund et al., 2017; Sadoyama and Miyano, 1981; Small, McNaughton, Greig, Lohkamp, & Lovell, 2009; Verrall, et al., 2005; Wollin, et al., 2017). Additionally, muscle fatigue has complex interactions with motor control and muscle functioning, presenting further mechanisms with regards to the role of fatigue and HSI incidences (Huygaerts, et al., 2020).

Fatigue also appears to have a greater negative effect upon both isometric and eccentric measures of strength in comparison to concentric measures (Carmona et al., 2018; Greig, 2008; Kilduff et al., 2008; Lord, Blazevich, et al., 2018; Lord, Ma'ayah, et al., 2018; Lovell, Siegler, Knox, Brennan, & Marshall, 2016; Pinto, Blazevich, Andersen, Mil-Homens, & Pinto, 2018). This in an important consideration, as if eccentric strength has greater influence of HSI incidence as described (Opar, et al., 2012), then it would be expected that the risk of injury is heightened under fatigued conditions. Although physical characteristics, which could minimise match play/training fatigue, i.e. VO_2 max, Yo-Yo intermittent fitness assessment and 40 m sprint, are not related to HSI incidence (Green, et al., 2020), therefore the influence of fatigue could be more of a local muscular factor. Fatigue is an influential risk factor for HSI incidence, however, it also likely to be closely related to overall hamstrings force generating capacity to able to cope with the demands of match play and training.

2.4.2.5 Flexibility

Flexibility, and improvements in flexibility from training have been proposed to reduce the risk of injury due to passive components of the MTU having greater compliance, allowing for a greater absorption of energy (Arnason, Andersen, Holme, Engebretsen, & Bahr, 2008; Opar, et al., 2012), although this is disputed within the research (van Beijsterveldt, van de Port, Vereijken, & Backx, 2013). Moreover, a high intensity stretching intervention has been observed to increase BF_{LH} FL aiding in preventing HSI risk (Freitas and Mil-Homens, 2015). To date, prospective studies has demonstrated that measures of flexibility have no association with the risk of HSI within American footballers (Burkett, 1970), AFL players (Bennell, Tully, & Harvey, 1999; Gabbe, Bennell, Finch, et al., 2006; Orchard, et al., 1997), soccer players (van doormaal, van der Horst, & Backx, 2016), or sprinters (Yeung, et al., 2009). Contrastingly, reduced hamstring flexibility in elite soccer players has been shown to be a significant risk factor in the occurrence of HSI (Witvrouw, Mahieu, Danneels, & McNair, 2004). Additionally, a hamstring stretching intervention resulted in a decrease in HSIs sustained during 13-weeks of military basic infantry training (Hartig and Henderson, 1999), although it could be suggested that the mechanisms of HSI within military basic infantry training would be different involving less high-speed running and more crouched or hip flexed running particularly with additional load. A recent meta-analysis identified that there were no clear relationships with HSI risk and any measure of hamstring flexibility, mobility and range of motion (Green, et al., 2020), although there is some conflicting evidence for hip extension and ankle dorsiflexion being weak risk factors. It should be noted that there are numerous methodological concerns with the measures of flexibility that should not be ignored. These include the accuracy and ecological validity of tests involved (Opar, et al., 2012), with no gold standard measurement for hamstring flexibility being established (Foreman et al., 2006).

2.5 Systematic Review and Meta-Analysis of Hamstring Strain Injury Prevention Practices and the Effect of Practice Compliance

The alarming incidence and cost statistics demonstrate the need to intervene, with appropriately designed training interventions that have the ability to reduce the occurrence of HSIs. Researchers have previously identified that the implementation of strength training, with an eccentric bias has the ability to reduce the risk of future HSI occurrence (Askling, Karlsson, & Thorstensson, 2003a; Mjølsnes, Arnason, Østhagen, Raastad, & Bahr, 2004; Petersen, Thorborg, Nielsen, Budtz-Jorgensen, & Holmich, 2011; Seagrave et al., 2014). However, Bourne et al.(2018) highlighted that the resultant risk reducing benefits only occur when an adequate intervention compliance is achieved, although what is an adequate level of compliance? A key issue within elite sport currently is that evidence-based hamstring injury prevention exercise, namely the NHE, is continually not being adopted by many champions league or Norwegian premier league clubs (Bahr, et al., 2015). A common complaint by players and coaches being that of delayed onset muscle soreness (DOMS) from the eccentric nature of the NHE (Morgan, 1990; Morgan and Proske, 2004). Hence, a training intervention that facilitates a wider scale adoption, which reduces the incidence of HSIs requires exploration.

With the global aim of reducing HSI events within sport, several preventative methods have been proposed. The two most common practices involve the implementation of a hamstring eccentric strength exercise (Petersen, et al., 2011), as well as the practice of a specific warm up protocol with the aim of reducing lower extremity injury, namely the FIFA 11 and FIFA 11+ (Silvers-Granelli et al., 2015; Thorborg, et al., 2017). Although both the FIFA 11 and FIFA 11+ warm up protocols include a hamstring specific eccentric exercise, specifically the NHE (Sadigursky et al., 2017). Both risk-reducing practices have previously been examined in a meta-analysis format (Goode, et al., 2015; Thorborg, et al., 2017; van Dyk, et al., 2019). All these meta-analyses demonstrate that eccentric resistance training and the FIFA 11+ have the potential to decrease the occurrence of hamstring injury in athletic populations (Goode, et al., 2015; Thorborg, et al., 2017), with up to a 50% reduction in HSI occurrence (van Dyk, et al., 2019). However, they continually report that the adoption and implementation appears adequate at best, with intervention compliance being a key component for an effective eccentric resistance training (Goode, et al., 2015). Similarly, for the FIFA 11+, less than 15% of intervention teams completed the recommended volume, as such this compromises the risk-reducing effectiveness of the FIFA 11+, in addition to the resultant risk ratios reported within the meta-analysis (Thorborg, et al., 2017). Goode et al.(2015) identified that with increased compliance there was a 65% decrease risk of HSI occurrence, however no study to date has looked to quantify what an adequate level of compliance is, for an intervention to be deemed effective. Grouping of studies in accordance with compliance to the intervention has been used previously, van Reijen et al. (2016) differentiated studies by <24.7%, 24.8-48.1% and >48.2%. However, given the importance of reducing HSI in athletic populations (Ekstrand, 2013; Ekstrand, Walden, et al., 2016), a greater compliance should be aimed for in HSI prevention interventions. Therefore, novel thresholds require identification for practitioners, as a guide to appropriate training practices.

The effect that interventions have on HSI occurrence has previously been examined within the literature (Goode, et al., 2015; Thorborg, et al., 2017). However, to date, observing the effect of intervention compliance on HSI risk has never been performed, despite commentary that achieving a high level of intervention compliance is crucial in reducing injury risk (Bourne, et al., 2018). Therefore, the purpose of this review is twofold: (1) to systematically review randomised control trials (RCT) examining the effects of HSI prevention programmes that hypothesised increases in strength of the hamstrings or associated structures, on the prevention of HSIs among athletes, and (2) to quantitatively explore the effect of intervention compliance on HSI injury risk.

2.5.1 Methods

2.5.1.1 Study design

The design of this systematic review was developed using the guidelines of the *Preferred Reporting Items for Systematic Reviews and Meta-Analyses* (PRISMA). The PRISMA statement includes a 27-item checklist that is designed to be used as a basis for reporting systematic review of randomised trials (Moher, Liberati, Tetzlaff, Altman, & Group, 2009). A review protocol was not registered for this review.

2.5.1.2 Search Strategy

A systematic, computerised search of the literature in PubMed, SPORTDiscus, MEDLINE, Scopus and Web of science was conducted, with controlled vocabulary and key words related to hamstring injury prevention programmes and hamstring injury. Our search timeframe was from inception to 08 May 2019. Key words (Table 2-1) were chosen in accordance with the aims of the research. Search terms were combined by Boolean logic (AND [between categories], OR [within categories]). Reference lists were also hand searched for any possible relevant studies.

Table 2-1 Summary of keyword grouping employed during database searches.

Injury	**Prevention**	**Training**	**Study**
Hamstring strain injury	Injury prevention	Resistance training	Randomised control trial
Hamstring injury	Hamstring injury prevention	Strength training	RCT
Posterior thigh injury	Primary prevention	Eccentric	Sport
Lower extremity strain	Injury prevention programmes	Eccentric training	Team sport
Lower limb injury	Injury risk reduction	Nordic hamstring exercise	soccer
	Compliance	Nordics	
		Russian curl	
		Warm up	
		FIFA 11	
		FIFA 11+	
		Plyometrics	
		Sprinting	

2.5.1.3 Selection criteria

All articles examining injury prevention programmes for the hamstrings were eligible for full-text review. An article was eligible for study inclusion if it met all of the following criteria: (1) the article was a RCT, (2) included athletes (participation in organised sports) of either sex who were at risk of incurring hamstring injuries and not participating in a hamstring rehabilitation programme, (3) included an intervention in comparison with a control or alternative intervention for the prevention of HSI, (4) interventions that aimed to increase strength of the hamstrings or associated structures. An article was excluded if: (1) included athletes with existing, or under treatment for, lower-limb musculoskeletal injuries, (2) reports focused on children below the age of 10 years, or (3) the article was not in English. All criteria were independently applied by the lead author (NJR) to the full text of the articles that passed the eligibility screening of titles and abstracts.

2.5.1.4 Quality assessment

The methodological quality of individual studies was assessed using the Physiotherapy Evidence Database (PEDro) scale (*http://www.pedro.fhs.usyd.edu.au.*, no date). Results from individual study analysis of quality were used to identify common areas of methodological weaknesses across studies.

PEDro uses 11 criteria, and reviewed studies were awarded one point for each criterion that was clearly satisfied, for a potential maximum value of 10 points. Criteria included; (1) eligibility criteria reported; (2) random assignment; (3) concealed allocation; (4) groups similar at baseline regarding most important prognostic indicator; (5) blinding of participants; (6) blinding of therapists who administered the therapy; (7) blinding of assessors who measured key outcome; (8) measures of at least one key outcome were obtained from more than 85% of initial participants; (9) all participants received treatment or control condition as allocated; (10) results of between-group arithmetical comparisons are reported and (11) study provides point measures and measures of variability for at least one key outcome.

2.5.1.5 Statistical analyses

Data, including counts and description of methods were extracted manually from included studies. DerSimonian and Laird (1986) random effects models were used for all analyses (meta-analyses and sub-group), to produce summary log odds ratios (LOR), 95% confidence intervals (CIs). The weighted means difference percentage (WMD%) was calculated to represent the aggregated differences of each individual study weighted by their sample size, WMD% and the size of each plot are proportional to their sample size. Overall effects were identified and the test for overall effect identified via the Z statistic. We used this model to be consistent with previously reported reviews on the same outcome (Goldman and Jones, 2010; Goode, et al., 2015).

To observe the effect of compliance upon HSI risk, selected articles were grouped via the following thresholds of compliance: very high (>75.1%), moderate-high (50.1-75%), low-moderate (25.1-50%) and very low (<25%). Group analyses included LORs, 95% CI's and heterogeneity between intervention compliance groupings. Additionally, group analyses were also performed upon intervention type (eccentric resistance training, FIFA 11/FIFA 11+ and bounding).

Heterogeneity test statistics and their *p* values were used to assess consistency of reported LORs across studies and between interventions. I-squared statistic (I^2) were used to describe the percentage of total variation across studies due to heterogeneity rather than chance alone with values >50% to indicate substantial heterogeneity. Significant heterogeneity was indicated with a $p < 0.10$. A higher *p* value was chosen to test for heterogeneity since these tests have low power particularly where there are few studies analysed. The τ^2 is reported to describe the pooled among-study variance of true effects, thereby reflecting the magnitude of heterogeneity.

Publication bias was evaluated by funnel plots, Egger's test and fail-safe *N* using the Rosenthal method (Egger, Smith, Schneider, & Minder, 1997). A fail-safe number of effects was calculated to determine how many un-retrieved null effects would be needed to diminish the significance of the observed effects to $P < 0.05$. All analyses were conducted by one of the authors using Jamovi (Jamovi project (2018) Computer Software, Retrieved from https://www.jamovi.org).

2.5.2 Results

2.5.2.1 Search Results

Eight hundred and sixty-six titles were identified through database and reference searches. Thirty-two full text articles were assessed for eligibility for inclusion, resulting in twenty studies being excluded based on study design and patient type, and a single study that was redacted by the journal. The process of study selection and the number of studies excluded at each stage, with reasons for exclusion is available in Figure 2-3.

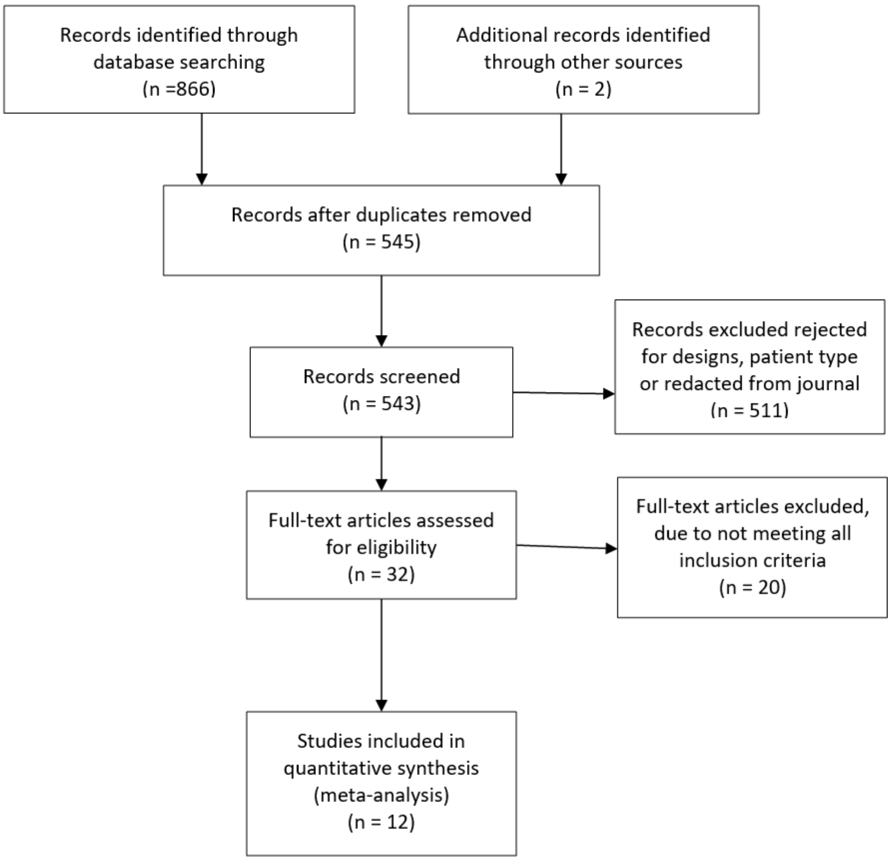


Figure 2-3 Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) flow diagram for study inclusion.

2.5.2.2 Characteristics of the included studies

The number of athletes in the studies ranged from 30 (Askling, et al., 2003a) to 1892 (Soligard et al., 2009). A description of the included studies' athlete populations, interventions, outcome measures, observation period and compliance are presented in Table 2-2.

Table 2-2 Summary of athletes, interventions, comparators, percentage compliance and injuries of included studies.

Reference	Athletes	Intervention Type	Intervention description	Observation period	*n* intervention	*n* control	Number of injuries intervention	Number of injuries control	Compliance (%)
Askling, et al. (2003a)	Male Swedish soccer players	Eccentric	10-week eccentric training intervention ("Yo-Yo" ergometer)	46 weeks	15	15	3	10	100
Engebretsen, et al. (2008)	Male Norwegian soccer players	Eccentric	10-week NHE training intervention from Mjølsnes et al. (2004)	1 season	85	76	23	17	21.1
Petersen, et al. (2011)	Male Danish soccer players	Eccentric	10-week NHE training intervention from Mjølsnes et al.(2004)	12 months	461	481	15	52	91
van der Horst, et al. (2015)	Male Dutch amateur soccer players	Eccentric	13-week NHE training intervention from Mjølsnes et al.(2004)	12 months	292	287	6	18	91
Gabbe, Branson, et al. (2006)	Male Australian amateur soccer players	Eccentric	NHE training intervention (12x6)	1 season	114	106	10	8	47
Sebelien, Stiller, Maher, & Qu (2014)	Male Semi-professional soccer players	Eccentric	Progressive NHE intervention, commencing with 2 x 5 once per week increasing to 3 x 8-12 three times per week.	1 season	59	60	0	6	22.7
del Ama Espinosa et al. (2015)	Female Elite European soccer players	Eccentric	NHE training intervention, 1 x 5 performed once per week for 42 weeks.	1 season	22	21	3	6	80
Saleh W (2017)	Male Australian amateur soccer players	FIFA 11+	Additional FIFA 11+ programme performed post-exercise two-three times per week.	6 months	144	136	2	9	83
Silvers-Granelli, et al. (2015)	Male NCAA collegiate athletes	FIFA 11+	FIFA 11+ three times per week.	5 months (August - December)	675	850	16	55	47
Van Beijsterveldt et al. (2012)	Male Dutch amateur soccer players	FIFA 11	FIFA 11 warm up twice per week.	1 season (9 months)	223	233	18.4% [38]	13.4% [29]	73
Soligard, et al. (2009)	Youth female soccer players (13-17 years).	FIFA 11+	FIFA 11+ warm up twice per week.	2 seasons (9 months)	1055	837	5	8	59.4
Van De Hoef et al. (2019)	Male Dutch amateur soccer players	Bounding	Bounding exercise programme.	1 season (9 months)	229	171	31	26	71

2.5.2.3 Quality of studies

The scores on each of the 11 criteria and total scores for each study are presented in Table 2-3. With quality assessment scores ranging from 4/11 to 7/11 for the highest scoring study (Petersen, et al., 2011).

Table 2-3 The PEDro quality assessment of individual studies.

Reference	1*	2	3	4	5	6	7	8	9	10	11	Total Score
Askling, et al. (2003a)	**-**	1	-	1	-	-	1	1	1	1	1	7
Engebretsen, et al. (2008)	**1**	1	-	1	-	-	-	1	1	1	1	6
Petersen, et al. (2011)	**1**	1	1	1	-	-	-	1	1	1	1	7
van der Horst, et al. (2015)	**1**	1	1	1	-	-	-	1	1	1	1	7
Gabbe, Branson, et al. (2006)	**1**	1	-	1	-	-	-	-	-	1	1	4
Sebelien, et al. (2014)	**1**	1	1	-	-	-	-	1	1	1	1	6
del Ama Espinosa, et al. (2015)	**1**	-	1	-	-	-	1	1	-	1	1	6
Saleh W (2017)	**1**	1	1	-	1	-	-	1	1	-	1	6
Silvers-Granelli, et al. (2015)	**1**	1	1	-	-	-	1	1	-	1	1	6
Van Beijsterveldt, et al. (2012)	**1**	1	1	1	-	-	-	1	-	-	1	5
Soligard, et al. (2009)	**1**	1	1	-	-	-	1	-	-	1	1	5
Van De Hoef, et al. (2019)	**1**	1	-	1	-	-	-	-	1	1	1	5

1. Eligibility criteria were specified. * Does not count to total score.
2. Subjects were randomly allocated to groups.
3. Allocation was concealed.
4. Groups were similar at baseline regarding most important prognostic indicators.
5. Blinding of all participants.
6. Blinding of coaches who administered the intervention.
7. Blinding of all assessors who measured at least one key therapy.
8. Measures of at least one key outcome obtained from more than 85% of the participants.
9. All subjects for whom outcome measures were available received the treatment or control condition as allocated.
10. Results of between-group statistical comparisons are reported for at least one key outcome.
11. Study provides both point measures of variability for at least one key outcome.
1, met criteria; -, criteria not met.
PEDRO, Physiotherapy Evidence Database.

2.5.2.4 Meta-analysis findings

Individual study LOR are illustrated in figure 2-4. This figure represents each individual study's LOR, 95% CI and WMD% of hamstring injury following the implementation prevention protocol. The diamond represents the overall summary estimate, and the width of the diamond represents the overall point estimate 95% CI. The overall pooled estimate from the main effects analysis was -0.64 (95% CI -1.10 to -0.17). The test for overall effect favoured the intervention treatments (Z = -2.70, p = 0.007). Significant heterogeneity was found between all studies (τ^2 = 0.405 (standard error = 0.280), I^2= 70.19%, p = < 0.001).

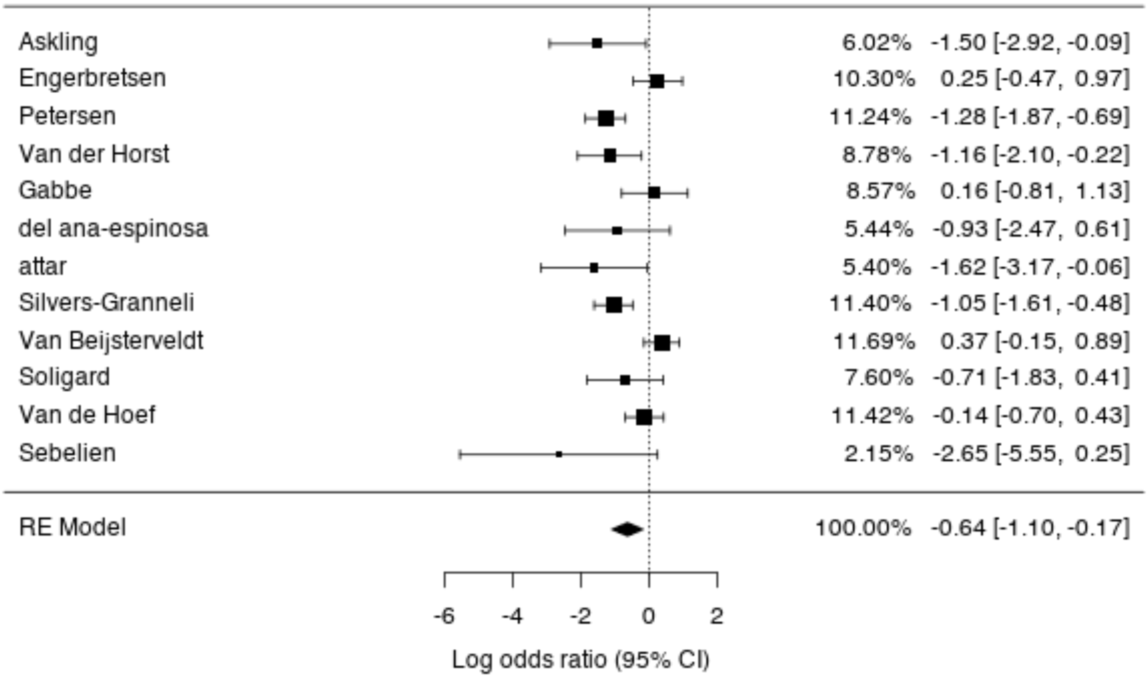


Figure 2-4 Forrest plot of individual study estimates and 95% CI, an overall pooled estimate (diamond) and 95% CI.

The effect of intervention compliance on LOR, 95% CI's and WMD% are demonstrated in figure 2-5, compliance was split into four sub-groups: very high compliance >75.1%, moderate-high compliance 50.1-75%, low-moderate compliance 25.1 - 50% and very low <25%. Within the current review 5/12 studies reported very high compliance (Askling, Karlsson and Thorstensson, 2003; J Petersen *et al.*, 2011; Horst *et al.*, 2015; Saleh *et al.*, 2017) (Askling, et al., 2003a; Petersen, et al., 2011; Saleh W, 2017; van der Horst, et al., 2015), 3/12 studies reported moderate-high compliance (Soligard, et al., 2009; Van Beijsterveldt, et al., 2012; Van De Hoef, et al., 2019), 2/12 studies reported low-moderate compliance (Engebretsen, et al., 2008; Silvers-Granelli, et al., 2015), and a 1/12 studies reported very low compliance. A significant difference was demonstrated between all levels of compliance (p < 0.001). With a meaningful trend of increased intervention effectiveness can be observed with increased compliance, with both very high- and moderate-high-compliance interventions being more effective than both low-moderate- and very low-compliance.

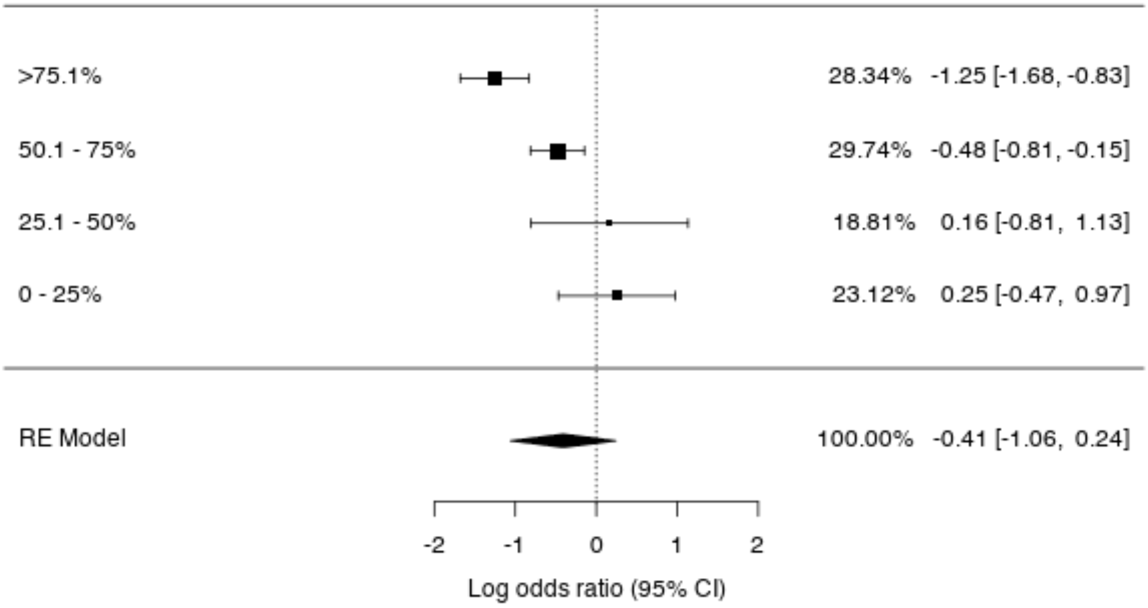


Figure 2-5 Forrest plot of intervention compliance grouped study estimates and 95% CI.

Figure 2-6 illustrates the pooled effects between intervention modality LOR, 95% CI's and WMD%, representing the effect of each type of prevention protocol on the probability of a HSI following the implementation of an intervention. No significant difference was demonstrated between intervention modalities (p = 0.199). However, eccentric exercise and FIFA 11/FIFA 11+ interventions appear more effective than bounding as an intervention modality.

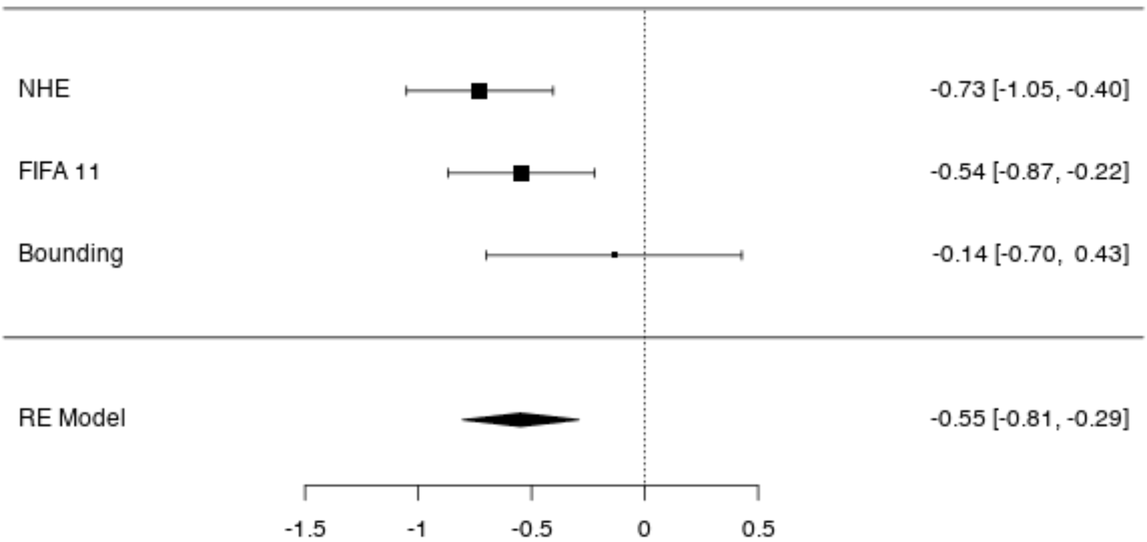


Figure 2-6 Forrest plot of intervention modality grouped study estimates and 95% CI.

2.5.2.4.1 Bias Assessment

The results of the Egger's test suggest that the mean effect of HSI risk reduction interventions within the present meta-analysis are subject to publication bias ($p < 0.001$). A funnel plot was used to visually assess symmetry and identify potential outliers (figure 2-7).

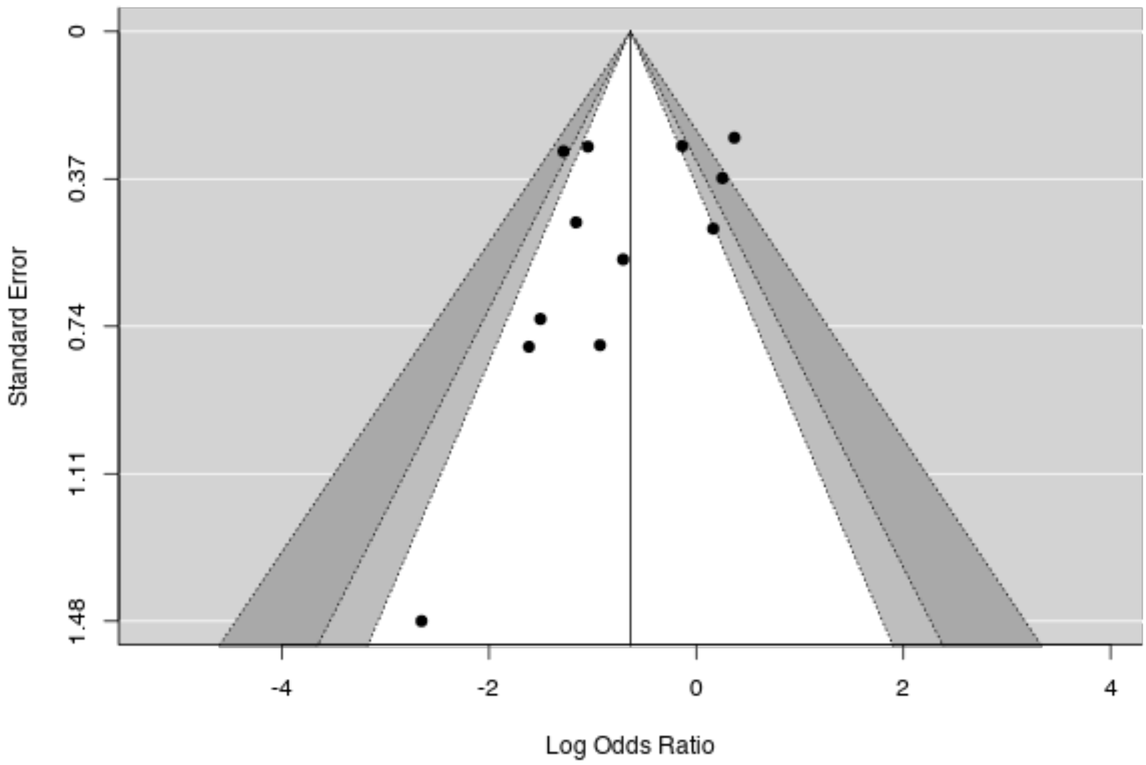


Figure 2-7 Funnel plot showing the publication bias results of the included studies.

2.5.3 Discussion

Within this systematic review on preventative measures of future HSIs, the literature was assessed with the effect of compliance and intervention type on the strength and direction of pooled study estimates were explored. Our search yielded 12 studies that met our inclusion criteria. Using data from these sources found similar preventative effects towards HSI prevention as reported by previous research (Goldman and Jones, 2010; Goode, et al., 2015; Thorborg, et al., 2017). The data from the present study highlights that for prevention measures to have a positive effect upon HSI occurrence, a compliance of at least 50.1% should be achieved. Furthermore, with increased compliance (>75.1%) there is a 160% increase in intervention effectiveness. This provides novel information surrounding the level of compliance that should be achieved by practitioners when implementing such interventions. Additionally, statistically significant preventative effects were observed for eccentric training, incorporating the NHE, and the implementation of the FIFA 11+ (Thorborg, et al., 2017), whereas no significant preventative effect was observed for the bounding exercise programme.

When accounting for compliance it was identified that a greater intervention compliance (>50.1%) is crucial in effectiveness of hamstring injury prevention strategies. Furthermore, increases in intervention compliance (>75.1%) resulted in a greater preventative effect, with an increased effectiveness of 160% with very high compliance, when compared to high-moderate compliance (LOR: -1.25 Vs -0.48). Goode et al. (2015) performed an intention-to-treat analysis to observe the effect of intervention compliance on hamstring injuries where they demonstrated that when non-compliers were removed from the analysis there was a substantial (65%) decrease in the risk of future HSI from eccentric training. A similar 65%

reduction was observed in an observational intervention study following an eccentric NHE intervention study (Arnason, et al., 2008).

Common barriers to non-compliance in strength and conditioning and physiotherapy practices include: DOMS (Mjølsnes, et al., 2004), pain during exercise (Chan, Lonsdale, Ho, Yung, & Chan, 2009; Friedrich, Gittler, Halberstadt, Cermak, & Heiller, 1998; Hayden, van Tulder, & Tomlinson, 2005; Jack, Mclean, Moffett, & Gardiner, 2010), confusion regarding correct exercise execution (Jack, et al., 2010), and poor coach support (Jack, et al., 2010). Consistent with a previous review (Goode, et al., 2015), DOMS was reported to be a main reason for non-compliance across a number of studies that were included within this review (Engebretsen, et al., 2008; Seagrave, et al., 2014; Soligard, et al., 2009; van der Horst, et al., 2015). Gabbe et al. (2006) identified that athletes may believe that DOMS increases their risk of future HSI, which would likely affect intervention compliance. Furthermore, the high volume of eccentric hamstring exercise prescribed within the interventions (Askling, et al., 2003a; Engebretsen, et al., 2008; Silvers-Granelli, et al., 2015; Soligard, et al., 2009; Van Beijsterveldt, et al., 2012; van der Horst, et al., 2015), could be a contributing factor in resultant DOMS and non-compliance (Goode, et al., 2015). More recently low volumes of the NHE have been shown to result in similar positive training adaptations which may contribute to the reduction in future HSI occurrence (Presland, Timmins, Bourne, Williams, & Opar, 2018). Furthermore, as the magnitude of the repeated bout effect is similar between high and low volumes of eccentric exercise (Howatson and van Someren, 2007); the potential positive effects of low volume NHE training on HSI incidence could be hypothesised. Due to the similarity in repeated bout effect between eccentric exercise volumes, if eccentric volume is decreased there would be a decrease in muscle damage and thus resultant muscle soreness, but adaptation would still occur (Howatson and van Someren, 2007; M. P. McHugh, Tyler, Greenberg, & Gleim, 2002; Nosaka, Lavender, Newton, & Sacco, 2003). This indicates that intervention compliance maybe improved upon by the implementation of low volume eccentric hamstring exercises, as there would be a reduction in ensuing DOMS. A prospective cohort study by Seagrave et al. (2014) identified a critical minimum volume of the NHE being 3.5 repetitions per week may reduce the occurrence of HSIs within professional baseball players when compared to a control group, however DOMS was still reported as major reason for non-compliance. One possible explanation for this non-compliance could be that the critical volume was the average number of completed repetitions across the season with no standardization or structured programming, which may have resulted in several weeks of detraining followed by a single high-volume week resulting in a high degree of DOMS.

Athlete boredom and motivation were further identified as barriers to non-compliance to interventions (Engebretsen, et al., 2008). One possible method of overcoming this maybe by providing direct supervision by trained professionals, who can offer encouragement and support (Goode, et al., 2015). Additionally, the use of novel devices that can provide real-time augmented feedback to the performance of tasks, such as the NHE, has the potential to increase athlete exertion (i.e., increased mean eccentric force (Chalker, Shield, Opar, Rathbone, & Keogh, 2018)), thus the potential for increased adaptive response and HSI preventative effect. Four studies included within this review reported providing direct supervision to athletes for the duration of the study (Askling, et al., 2003a; Petersen, et al., 2011; Seagrave, et al., 2014; Soligard, et al., 2009). Although the quality and reported compliance varied between the studies, the effect of regular and consistent feedback received from: sports coaches, strength and conditioning coaches, physical therapists, physicians, or peers, should not be underestimated in the role for a positive change. Although on-field supervision of the FIFA 11+ warm up intervention demonstrated only a minimal effect on performance of the intervention (Steffen, Emery, et al., 2013; Steffen, Meeuwisse, et al., 2013), there was a substantial difference in the volume of exercises performed (Steffen, Emery, et al., 2013; Steffen, Meeuwisse, et al., 2013). Moreover, direct supervision could improve exercise quality, thus improving intervention effectiveness (Goode, et al., 2015). Additionally, improving athlete and coach education will aid in debunking common beliefs, including that performing eccentric exercises may increase the risk of future HSIs (Gabbe, Branson, et al., 2006), in addition to providing a greater understanding of the preventative value of their implementation (Steffen, Emery, et al., 2013; Steffen, Meeuwisse, et al., 2013), improving intervention compliance. Non-compliance of HSI interventions could be tackled with the modification of training protocols to utilise a low volume approach to eccentric strengthening (Presland, et al., 2018), while providing feedback and support through direct supervision. Future research should therefore be directed to the potential of low volume of eccentric strengthening exercises with an interest in intervention compliance, as well as the potential of implementing other intervention protocols that may achieve greater athlete compliance e.g. sprint based interventions (Freeman et al., 2019).

Intervention type on future HSI was also observed within the present study (figure 8). The interventions were split across three types which all aimed to increase strength of the hamstrings and/or associated structures, including eccentric exercise, FIFA 11/FIFA11+ warmups and bounding training interventions. The pooled effects demonstrated that two intervention types (i.e., eccentric exercise and FIFA 11/FIFA11+ warmups), decreased the occurrence of future HSI. While bounding, provided a minimal decrease in the risk of future HSI occurrence, as although the observed LORs are less than zero (-0.14), the 95%CIs include zero. Within a recent review and meta-analyses, eccentric hamstring training (i.e., NHE), has been found to decrease the risk of injury by up to 50% (van Dyk, et al., 2019). This is potentially as a result of the positive adaptations that have been shown to occur following their implementation including increased BF_{LH} FL and increased force production across muscle actions, joint angles and movement velocities (Ishoi et al., 2018; Mjølsnes, et al., 2004; Nosaka, et al., 2003; Presland, et al., 2018; Ribeiro-Alvares, Marques, Vaz, & Baroni, 2018). Similarly, within the current review, eccentric hamstring training had the greatest preventative effect, however there was only a minimal difference between the eccentric hamstring training and the warm-up interventions. An explanation for this minimal difference could be that the warmup interventions examined within this review (FIFA 11 and FIFA 11+), includes the NHE. No research to date has demonstrated what adaptations may occur from the implementation of the FIFA 11 and FIFA 11+ that may aid in hamstring injury prevention. Nevertheless, the warmup interventions still offer a positive effect on the risk of future HSI occurrence, making it an effective, practical and time efficient practice in sport. The bounding intervention, with the inclusion of dynamic lunges and bounding variations over incremental distances, implemented by van de Hoef et al. (2019) may not have elicited a desired preventative effect as hypothesised, as the magnitude of hamstring loading may have not been a sufficient stimulus for an adaptive response to occur, although no measure of strength or muscle architecture were observed (Van De Hoef, et al., 2019).

The majority of studies included in this review appear to favour the intervention (figure 2-4), when the aim is to reduce the occurrence of HSI within soccer athletes, however three studies within the current review appear to potentially favour the control (Engebretsen, et al., 2008; Gabbe, Branson, et al., 2006; Van Beijsterveldt, et al., 2012). Several possible explanations as to why this result was identified could be explored. Firstly, both Engebretsen et al. (2008) and Gabbe et al. (2006) implemented extremely high-volume protocols, Mjolsnes protocol (2004) and 12 sets of six (Gabbe, Branson, et al., 2006), respectively. These higher volume interventions can result in excessive fatigue and DOMS, with both factors having negative impact on compliance, as discussed earlier. Additionally, Engbretsen's et al. (2008) RCT design included soccer players who were deemed to be at a higher risk of future HSI, with high risk players reporting previous HSI occurrence. Previous research has demonstrated that the risk of future HSI incidence is greater within athletes who have a history of previous lower limb injury, including HSI, ankle, knee, calf, quadriceps, anterior cruciate ligament, chronic groin, lumbar stress fracture and severe back injuries (Malliaropoulos et al., 2018; Timmins, Bourne, et al., 2016; Toohey, Drew, Cook, Finch, & Gaida, 2017). Furthermore, the consistency of intervention application can also be questioned within the RCT study by Gabbe et al. (2006), as their protocol (12 x 6) was performed on total of five occasions within a 12-week period, whereby multiple weeks could pass prior to the subsequent dose. This becomes an issue as the structural and force producing capabilities of the hamstrings can rapidly return to baseline in as little of two-weeks, and therefore potentially losing their preventative adaptations. A possible explanation as to why Van Beijsterveldt and colleagues' study (2012) favoured a control over the warm-up intervention, is that it is the only study that utilised a FIFA 11 intervention compared to the FIFA 11+. This difference in intervention design is important as the FIFA 11 protocol has been shown to have a minimal effect on injury rates in footballers when compared to the FIFA 11+ intervention (Thorborg, et al., 2017). The FIFA 11+ protocol incorporates a number of supplementary exercises in comparison to the FIFA 11 protocol including: running drills and squats (Thorborg, et al., 2017). In addition to the supplementary exercises, the FIFA 11+ protocol has included several progressive levels where athletes are exposed to increased intensity and volume, which may have further enhanced the preventative effects (Thorborg, et al., 2017).

The current review is not without methodological limitations. Firstly, only one author was involved in the study selection process, however using similar search strategies that have been reported within previous searches as recent as August 2018 (Goode, et al., 2015; Thorborg, et al., 2017; van Dyk, et al., 2019). Using the previously reported search strategies, a similar volume of records was discovered, which eventually resulted in all articles which have been reported previously being discovered along with a novel interventions modality (i.e., bounding). Additionally, even though the searches were conducted across multiple databases, relevant studies could have been excluded as the search was limited to the English language. There may be no way to truly know the number of unpublished studies that exist in the "file drawer". A conservative estimate that for the 12 published studies identified in the current analysis, upwards of 92 unpublished and undiscovered studies may still be filed away (Rosenthal, 1979). Within the current review, effects were pooled into subgroups by intervention modality and intervention compliance, this is without the removal of possible study outliers identified by funnel plot (Engebretsen, et al., 2008; Van Beijsterveldt, et al., 2012), potentially impacting on the determined effects. However, the removal of study outliers would be contraindicated as both studies still offer an insight into HSI risk reduction strategies within sport and the possibility of null effects. Furthermore, the funnel-shaped plot (figure 2-7), illustrating the observed effects vs the standard error can be disrupted by the heterogeneity of the studies, thereby increasing the likelihood of false-negative and false-positive decisions about publication bias (Hopkins, 2018). Intention-to-treat analysis has been described as the preferred method of determining effectiveness of interventions in RCT (Goode, et al., 2015), yet can be subject to null-bias where substantial non-compliance is reported (Goode, et al., 2015). However, as intention-to-treat analysis has been performed previously within a similar review (Goode, et al., 2015), and the aim of this review was to observe the effect of total intervention compliance providing a novel scale of very high-, high-moderate- and low-moderate and very compliance on the observed effect, it was deemed unnecessary.

2.5.4 Conclusions

The implementation and overall effectiveness of interventions is related to the observed compliance, with a linear increase in compliance leading to increased effectiveness. Compliance of >50.1% demonstrated a significant positive effect on the occurrence of future HSI. Crucially, further increases in compliance (>75.1%), resulted in an 160% increase in preventative effect, this highlights the need for practitioners to design and implement interventions where a compliance of >75.1% is achievable. Furthermore, eccentric resistance training and the FIFA 11/FIFA11+ appear to have an influential role in successful prevention of future HSI. It should be noted that the inclusion of the NHE within the FIFA 11+ may explain its effectiveness at reducing the occurrence of future HSIs. A bounding intervention offered a limited positive impact on the occurrence of future hamstring injury, however only a single intervention has utilised this methodology previously and therefore requires further investigation.

2.6 Systematic review and meta-analysis of the Effect of Hamstring Training Interventions on Bicep Femoris Fascicle length

The architectural properties of the hamstrings have been described as extremely pliable to imposed training demands, being termed a "plastic" muscle (Bourne, et al., 2018; Timmins, Bourne, et al., 2016; Timmins, Ruddy, et al., 2016; Timmins, Shield, Williams, Lorenzen, et al., 2016). Researchers have identified that team sport athletes who had a previous HSI history possessed significantly shorter BH_{LH} fascicles when compared to athletes with no previous HSI history (Bourne, et al., 2018; Shield and Bourne, 2018; Timmins, Bourne, et al., 2016; Timmins, et al., 2015). Furthermore, the results of a number of resistance training interventions have demonstrated significant changes in BH_{LH} FL, with changes being training mode specific (Bourne, Duhig, et al., 2017; Bourne, et al., 2018; Pollard, Opar, Williams, Bourne, & Timmins, 2019; Presland, et al., 2018; Timmins, Ruddy, et al., 2016). However, to date no review has looked to consolidate the information regarding hamstring specific FL adaptations to training. Therefore, the aim of this review is to systematically review the literature across resistance-based training interventions which measured the changes on BF_{LH} FL utilising ultrasound methodologies.

2.6.1 Methods

A systematic, computerised search of the literature in PubMed, SPORTDiscus, MEDLINE, Scopus and Web of science was conducted, with controlled vocabulary and key words related to hamstring injury prevention programmes and hamstring injury. The search timeframe was from inception to April 2020. Key words (Table 2-4) were chosen in accordance with the aims of the research. Search terms were combined by Boolean logic (AND [between categories], OR [within categories]). We also extended the search spectrum to "related articles" and the bibliographies of all retrieved studies.

Table 2-4 Summary of keyword grouping employed during database searches.

Anatomy	Architecture	Imposed demand	Methodology
Hamstring	Fascicle length	Resistance training	Ultrasound
Bicep femoris	Pennation angle	Eccentric	
	Fascicle angle	Concentric	
	Muscle architecture	training	

2.6.1.1 Selection criteria

The following inclusion criteria were used to select articles focused on the effect of training and HSI on BF_{LH} architecture:

1. Full-text, research articles exploring and analysing adaptations BF_{LH} architecture were selected. As such, case studies, review articles, and abstracts were excluded.
2. Research articles must report BF_{LH} architecture characteristics (FL, PA and MT or volume) pre- and post-training or injury occurrence.

2.6.1.2 Quality assessment

Study quality was evaluated by a standard procedure (Table 2-5). Each study was read and ranked from 0 to 6, with the larger number indicating better quality. For each question a 1 was awarded if the study met the standard. If insufficient description or data were provided to analyse a specific question, 0 was awarded. The score was tallied for each question, with the highest score possible equalling 6 out of 6.

Table 2-5 Quality assessment of bicep femoris architecture intervention studies.

			Score
A	**Sample description:**		0 or 1
	+	Properties of the subjects (age, mass, height, sex)	
	+	Definition of the population (well-trained, recreationally trained, untrained)	
B	**Intervention:**		0 or 1
	+	Defined and supervised training programme (exercise performance, coaching, progressions)	
	+	Defined volume and frequency	
C	**Methods employed for bicep femoris long head architecture assessment:**		0 or 1
	+	- Ultrasound (probe length, scanning site, frequency, assessment position, number of images)	
D	**Data analysis:**		0 or 1
	+	Defined estimation or measurement methodology	
	+	Defined software for analysing data	
E	**Results detailed:**		0 or 1
	+	Measure of central tendency and variation from the average	
	+	Difference or magnitude of the difference provided	
F	**Reliability of assessor detailed:**		0 or 1
	+	Defined and developed reliability test, regarding ultrasound collection of the bicep femoris	

2.6.1.3 Statistical analyses

To assess the magnitude of each training stimulus, where possible Hedge's *g* effect sizes (ES) were calculated from the mean and standard deviation and sample size. Where the mean and standard deviation were not reported by the authors, with only magnitude-based differences described, typically Cohen's *d* ES, conversions to Hedge's *g* ES were made to allow comparison between studies and training modalities (Equation 2-1) (Lakens, 2013).

$$Hedge'sg = Cohen's\ d \times (1 - \left(\frac{3}{4 \times (n-1) - 1}\right))$$

Equation 2-1 Effect size conversion from Cohen's d to Hedge's g.

Alongside the magnitude of differences, the mean difference was also reported for each study. The scale for interpretation of ES was proposed by Hopkins (2010) as follows; trivial (≤ 0.20), small (0.20–0.59), moderate (0.60–1.19), large (1.20–1.99), or very large (≥2.00).

DerSimonian and Laird (1986) random effects models was used to observe the overall effect using the Z statistic. Consistency of effects was quantified using a test for heterogeneity (*I^2*) outlined by Higgins, Thompson, Deeks, & Altman (2003) whereby a scale of low (<25%), moderate (25-75%) and high (>75%) I^2 values were used for the interpretation of consistency. The duration of study intervention was used as a moderator within the DerSimonian and Laird (1986) random effects model, to observe the effect of study duration on the magnitude of adaptations.

2.6.2 Results

2.6.2.1 Search Results

A flow diagram of the literature search and the final selection is shown in Figure 2-8. According to the above-defined inclusion criteria, 16 independent studies were identified. Across the 12 studies, all identified BF_{LH} architecture measurement pre- and post-training intervention.

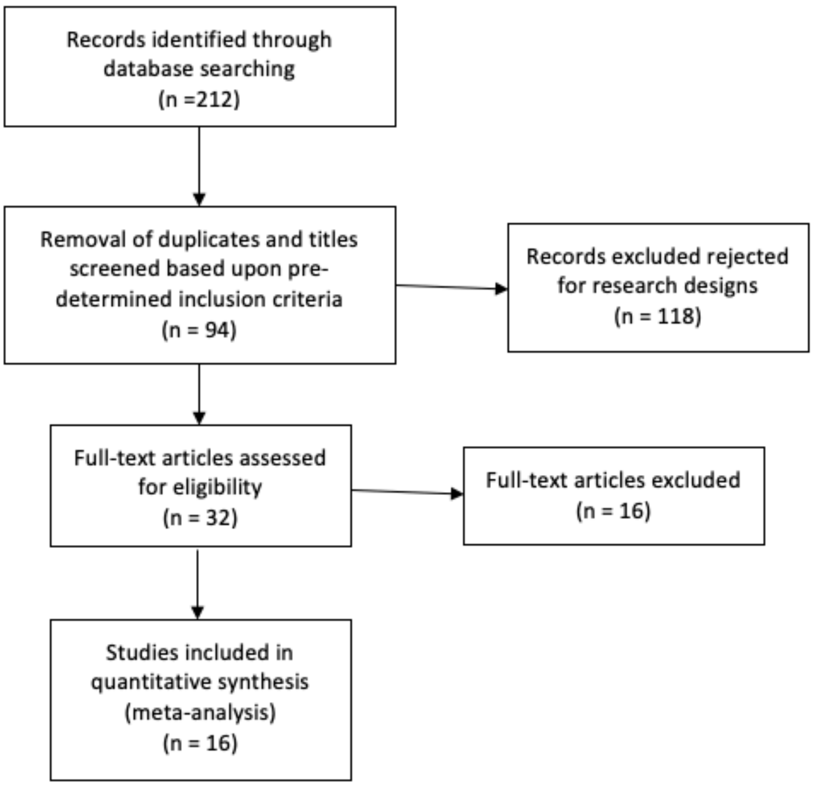


Figure 2-8 Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) flow diagram for study inclusion.

2.6.2.2 Study description

The characteristics of the studies are presented in Table 2-6.

Table 2-6 Individual training intervention study characteristics, descriptive, magnitude of change and quality assessment for Bicep femoris long head Fascicle length.

Year	Study	*n*	Participant's characteristics				Intervention prescription						Ultrasound methodology		Study quality	Fascicle length (cm)		Hedge's *g* (95% CI)
			Study population description	**Age (years)**	**Height (cm)**	**Mass (kg)**	**Length (Weeks)**	**Modality**	**Average weekly dose**	**Frequency**	**Volume per session**	**Intensity**	**Probe length (cm)**	**Method**		**PRE (SD)**	**POST (SD)**	
2009	**Potier**	11	Non-resistance trained	29.6 ± 1.2	168.0 ± 3.6	64.3 ± 1.2	8	Eccentric single leg curl	**NR**	**NR**	1 - 24	100% RM	4.1	Linear extrapolation	3	5.90 (0.30)	7.88 (0.35)	5.61 (2.78 - 8.39)
2016	**Timmins**	14	Recreationally active	22.3 ± 4.2	181.0 ± 7.0	76.9 ± 8.2	6	Concentric Isokinetic	97	2 - 3	24 - 48	No progression	4.7	FL estimation equation 1	6	11.71 (0.90)	10.33 (0.80)	1.53 (0.30 - 2.72)
		14						Eccentric Isokinetic								11.53 (0.60)	13.42 (0.80)	2.52 (1.05 - 3.94)
	Guex	11	Recreationally active	27.3 ± 3.9	173.5 ± 10.8	66.0 ± 13.6	3	SL eccentric isokinetic	90.6	2 - 3	24 - 40	No progression	4.2	Linear extrapolation	5	8.41 (0.73)	8.82 (0.79)	0.50 (-0.72 - 1.69)
		11		28.4 ± 4.5	170.7 ± 5.9	64.0 ± 12.7		LL eccentric isokinetic								8.20 (0.93)	8.94 (0.81)	0.78 (-0.47 - 1.99)
2017	**Seymore**	10	Recreationally active	18.3 ± 0.5	170.0 ± 10	71.3 ± 15.9	6	NHE	62.6	1 - 3	10 - 30	No progression	**NR**	Panoramic	6	8.96 (1.23)	9.07 (1.73)	0.07 (-1.17 - 1.31)
	Ribeiro-Alvares	10	Moderately active	26.0 ± 2.7	166.4 ± 7.2	63.7 ± 11.1	4	NHE	46.5	2	18 - 30	No progression	4.0	Linear extrapolation	5	8.36 (0.63)	10.18 (0.75)	2.40 (0.67 - 4.06)
	Bourne	10	Recreationally active	22.0 ± 3.6	180.4 ± 7.0	80.8 ± 11.1	10	single leg 45° hip extension	64.6 - 72.6	2	12 - 50	60-80% RM	4.7	FL estimation equation 1	6	**NR**	**NR**	1.62 (0.12 - 3.05)
		10						NHE				2.5kg Incremental loading				**NR**	**NR**	1.98 (0.38 - 3.51)
2018	**Alonso-Fernandez**	23	Recreationally active	25.2 ± 3.3	176 ± 9.0	75.5 ± 8.1	8	NHE	57.75	2 - 3	12 - 30	No progression	4.7	FL estimation equation 1	5	8.17 (1.83)	10.12 (1.85)	1.02 (0.14 - 1.88)
	Presland	10	Recreationally active	22.3 ± 2.8	179.1 ± 7.7	75.1 ± 8.8	6	NHE	21.3	1 - 2	8 - 48	2.5kg Incremental loading	4.7	FL estimation equation 1	6	10.09 (0.67)	12.50 (0.72)	3.17 (1.17 - 5.10)
		10						NHE	73.3	2	48 - 100					10.18 (0.66)	12.56 (0.97)	2.62 (0.81 - 4.35)

2019	**Duhig**	15	Recreationally active	22.8 ± 4.1	180.1 ± 6.4	85.2 ± 14.6	5	Concentric leg curl	39.6	1 - 2	12 - 30	6-7RM	4.7	FL estimation equation 1	6	10.39 (**NR**)	9.73 (**NR**)	0.87 (-0.27 - 1.92)
		15						NHE				5kg Incremental loading				10.22 (**NR**)	11.62 (**NR**)	1.89 (0.62 - 3.11)
	Pollard	10	Recreationally active	24 ± 4	181 ± 6	78 ± 11	6	Razor curl	21.3	1 - 2	8 - 48	2.5kg Incremental loading	4.7	FL estimation equation 1	6	9.76 (0.80)	9.61 (0.84)	0.17 (-1.08 - 1.41)
		10						BW NHE				No progression				9.85 (0.90)	10.61 (0.67)	0.88 (-0.45 - 2.71)
		10						Weighted NHE				2.5kg Incremental loading				9.85 (1.13)	11.67 (0.81)	1.69 (0.17 - 3.14)
	Lacome	10	Elite youth soccer players	17.5 ± 0.7	175.7 ± 5.0	64.7 ± 4.9	6+6 §	NHE and modified SLDL	40	1	40	No progression	4.2	Panoramic	6	8.30 (1.00)	8.70 (1.20)	0.33 (-1.00 - 1.64)
									10		10					8.70 (1.20)	8.70 (1.20)	0.00 (-1.24 - 1.24)
		9		17.2 ± 0.7	174.8 ± 6.1	64.1 ± 5.7			10		10					8.70 (1.50)	9.10 (1.20)	0.27 (-0.98 - 1.51)
									40		40					9.10 (1.40)	9.20 (1.20)	0.07 (-1.24 - 1.37)
2020	**Mendiguchia**	7	Elite Portuguese soccer players	**NR**	**NR**	**NR**	7	NHE	56.7 - 62.6	1 - 3	10 - 30	No progression	4.7	Linear extrapolation	5	9.93 (1.10)	10.66 (1.01)	0.60 (-0.95 - 2.10)
		8						Sprint	520 m	2	400 - 680 m	Sled - 15-70% BW Maximal effort sprints				10.23 (1.91)	11.89 (1.16)	0.93 (-0.58 - 2.38)

2020	**Presland**	10	Recreationally active	27.8 ± 5.3	178.4 ± 0.7	80.0 ± 10.70	6	Unilateral eccentric flywheel (0.1 kg·m)	65.3	2	24-48	No progression	4.7	FL estimation equation 1	6	9.51 (0.67)	10.88 (0.76)	1.81 (0.26 – 3.29)
		10						Unilateral conventional flywheel (0.05 kg·m)								9.64 (0.65)	9.61 (0.80)	-0.04 (-1.28 – 1.20)
	Marusic	18	Healthy individuals	23.4 ± 3.3	177.0 ± 7.0	78.0 ± 8.2	6	Long length eccentrics (GL and NHE)	37.3	1	20-48	5-10 kg progression on NHE, 8-20 kg progression on glider	NR	Panoramic	6	7.74 (0.82)	8.32 (0.85)	0.66 (-0.30 – 1.60)
	Medeiros	15	U20-23 professional footballers	18.8 ± 1.74	182 ± 0.08	78.8 ± 8.39	8	NHE once per week	28.5-30.5	1	12-40	No progression	4	FL estimation equation 1	6	10.02 (1.84)	10.93 (2.92)	0.35 (-0.68 – 1.36)
		17		18.5 ± 1.07	179 ± 0.10	75.5 ± 10.44		NHE twice per week	57-61	2						10.09 (2.12)	11.04 (2.74)	0.37 (-0.60 – 1.32)
Published ahead of print	**Severo-Silveira**	11	Competitive rugby players	27.2 ± 3.26	175 ± 0.05	90.1 ± 14.3	8	NHE constant volume	34.5	2	12-18	No progression	4	FL estimation equation 1	6	10.48 (2.74)	11.26 (2.83)	0.26 (-0.93 – 1.44)
		10		25.2 ± 3.34	176 ± 0.08	88.6 ± 12.8		NHE Progressive volume	57-61		12-40					11.13 (2.83)	12.26 (3.20)	0.34 (-0.92 – 1.58)

Trivial - <0.20	Small – 0.20 – 0.59	Moderate – 0.60 – 1.19	Large – 1.20 – 1.99	Very large - >2.00

NR = Not reported within the study, § = Crossover study design, SL = Short muscle length, LL = Long muscle length, NHE = Nordic hamstring exercise, SLDL = Stiff leg deadlift, GL = Glider, RM = One repetition maximum, BW = Bodyweight, FL estimation equation 1 = FL=sin (AA+90º)×MT÷sin (180º–(AA+180º–PA))

2.6.2.3 Hedge's g effect size identifying the magnitude of change in bicep femoris fascicle length

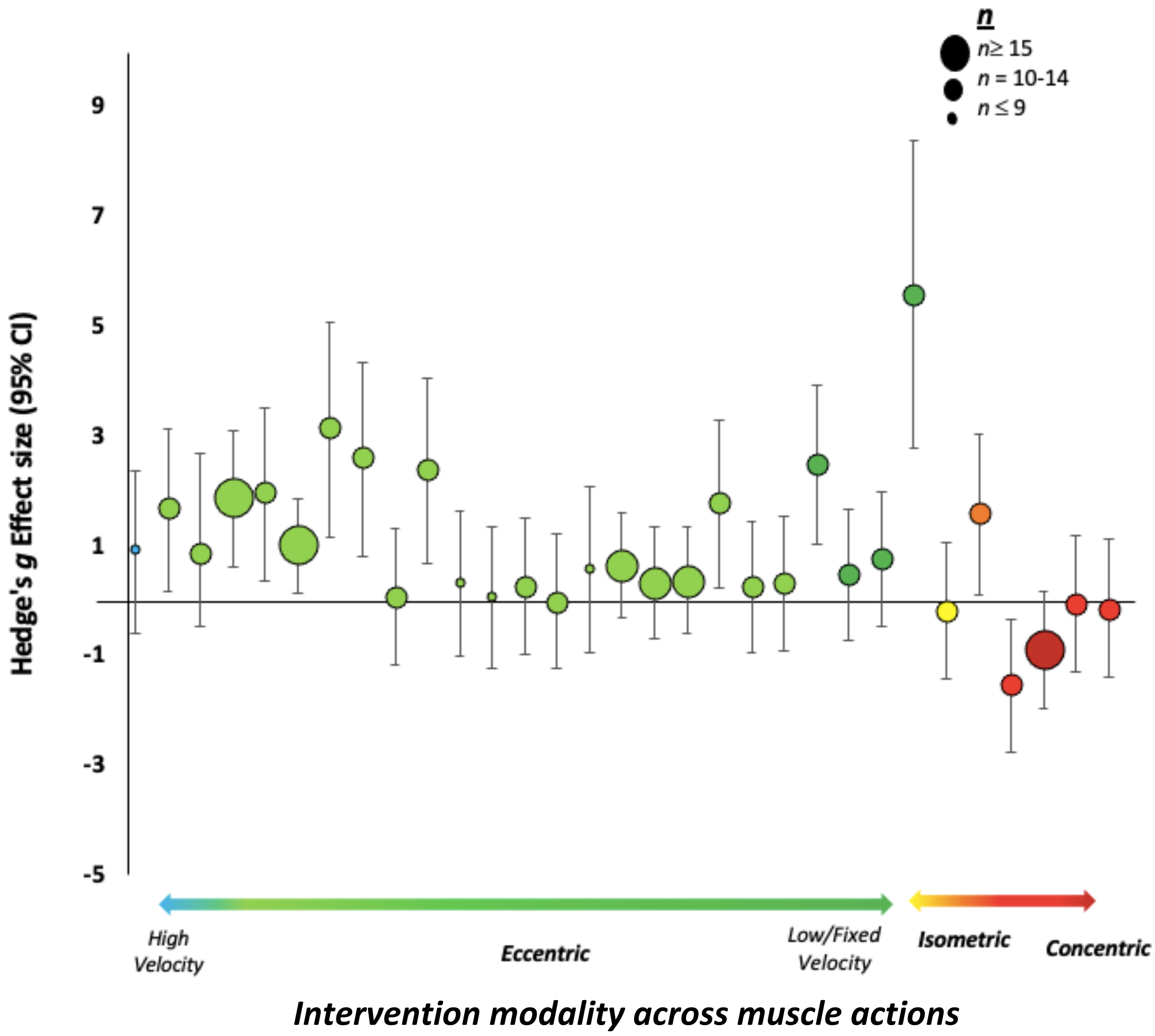


Figure 2-9 Magnitude of effects across training interventions, distinguishing primary muscle actions stimulus provided.

Along the spectrum of modalities, eccentric training typically resulted in an increase in BF_{LH} FL, isometric training resulted in minimal to no change in BF_{LH} FL, while concentric only modalities resulted in decreased BF_{LH} FL (Figure 2-9). Contrasting this trend however, a conventional concentric-eccentric resistance exercise saw a large positive increase in BF_{LH} FL. Although the magnitude of BF_{LH} FL adaptations were variable, with a number of 95% CI crossing the zero line (52%, 12/23 intervention groups).

The overall pooled estimate (Hedge's *g*) from the main effects analysis was 0.79 (95% CI 0.37 to 1.21), with the test for overall effect ($Z = 3.71$, $p < 0.001$). Low consistency was observed between the studies ($I^2 = 7.24\%$, $p = 0.352$). When using study duration as a moderator within the DerSimonian and Laird (1986) random effects model, there was a trivial-small estimate (0.09 (95% CI -0.29 to 0.46)), with the test for overall effect ($Z = 0.46$, $p = 0.643$) (Figure 2-10). Contrastingly, when using intervention modality as a moderator, there was a large estimate

(1.76 (95% CI -1.37 to 4.89) favouring eccentric modalities (supra-maximal and sprinting), with the test for overall effect (Z = 1.10, p = 0.270) (Figure 2-9 & 2-10).

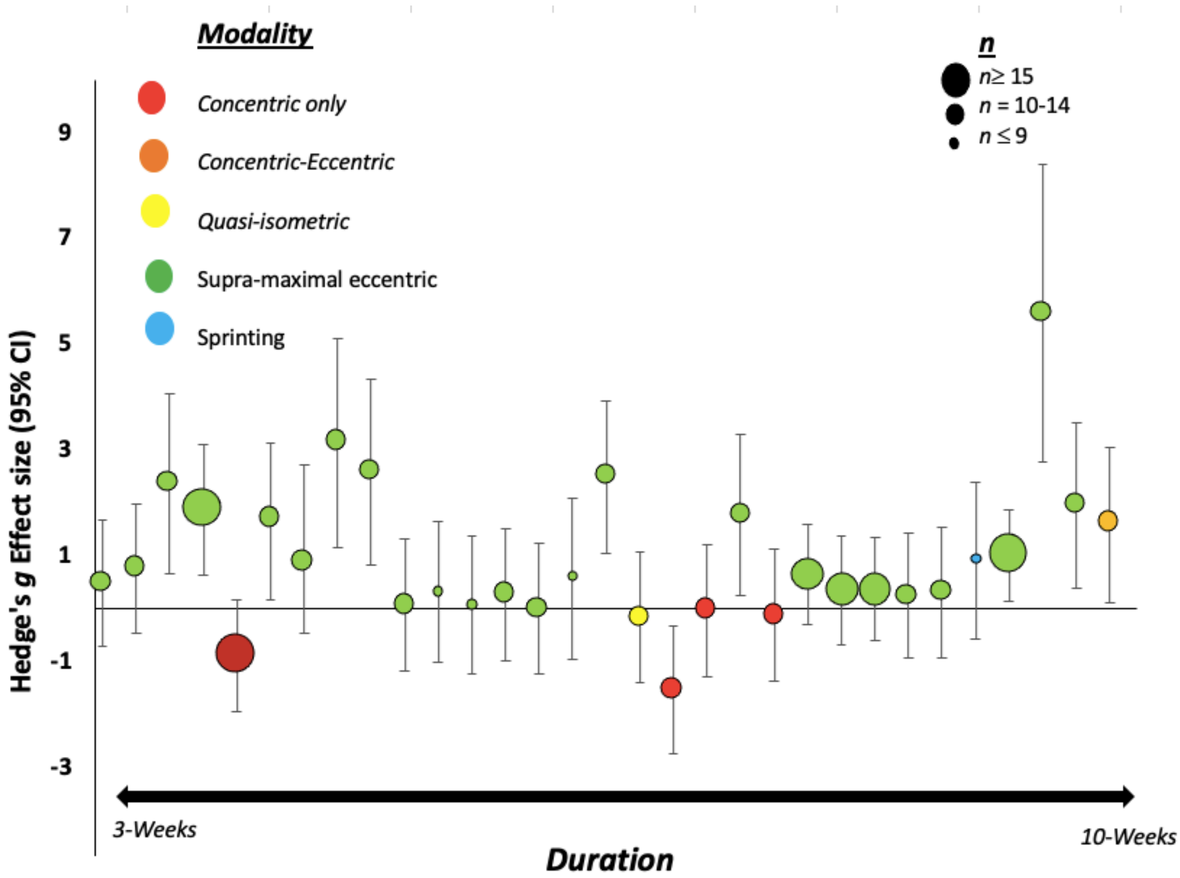


Figure 2-10 Magnitude of effects across training interventions, with study duration used as a moderator.

2.6.3 Discussion

The results of this review are in line with previous literature suggesting that architectural adaptations are mode specific (Timmins, Shield, Williams, Lorenzen, et al., 2016), with eccentric training stimuli typically increasing FL, albeit by varying magnitudes. Whereas a contrasting effect can be seen with concentric-only stimuli, where a decreased FL was observed. Interestingly, the incorporation of a submaximal eccentric (lengthening) component in a hip dominant concentric-eccentric task (single leg 45° hip extension) resulted in similar positive adaptations to BF_{LH} FL (Bourne, Duhig, et al., 2017). To date, a single study has been performed where the researchers observed the effect of a quasi-isometric exercise (razor curls) observing a trivial increase in BF_{LH} FL (Pollard, et al., 2019). Researchers recently reported the effect of a sprint training programme on architectural adaptations of the BF_{LH}, where a large increase in BF_{LH} FL was observed (Mendiguchia et al., 2020).

The largest observed effect was found for Potier, Alexander, & Seynnes (2009), this is the earliest study found within the present search criteria and the observed magnitude has a very straight forward explanation. The participant sample used within the study performed by Potier, et al. (2009), were non-athletic, potentially sedentary population where no previous history of resistance training was performed. This is further evidenced by the extremely low absolute BF_{LH} FLs reported at both Pre, and Post-intervention testing – with the lowest FLs reported for the BF_{LH} (5.90 cm) within the literature. This is remarkable as it is even lower than what is reported for cadaver specimens (Kellis, et al., 2012; Kellis, Galanis, Natsis, & Kapetanos, 2009; Kellis, et al., 2010). Therefore, no surprise that when provided with a progressive volume, maximal eccentric training stimulus a very large, positive increase in BF_{LH} FL was found (Potier, et al., 2009)**.**

With regards to the study quality assessment, the earliest study Potier, et al. (2009) was the lowest ranked included study (3/6), whereas all later studies were more highly rated, between 5-6. Potier, et al. (2009) main failure was that there was no specific description of specific training doses which prevents future study replication. Typically, studies which did not achieve the maximum study quality; failed to report reliability statistics. This information is crucial in all training intervention studies, as acknowledging the measurement error aids in understanding if any change is a meaningful adaptation as a result of the training stimulus or could just be random error within the measurement (Hachana, Attia, Nassib, Shephard, & Chelly, 2012; Hopkins, Marshall, Batterham, & Hanin, 2009; Swinton, Hemingway, Saunders, Gualano, & Dolan, 2018; Weir, 2005). This is especially true for muscle architecture, as the assessment and observed changes are a very subjective measurement with a number of sources of potential error (Franchi, Fitze, et al., 2019; Franchi, et al., 2018).

2.6.3.1 Eccentric interventions

Across the eccentric focused studies, two key modalities have been utilised, specifically isokinetic or alternative maximal-supramaximal exercises (e.g., NHE, flywheel or glider). Two studies have utilised eccentric isokinetic modes of action as part of training interventions, both interventions Guex et al, (2016) and Timmins et al, (2016) observed moderate to very large increases in FL. Timmins et al, (2016) found a very large increase in FL, whereas Guex et al, (2016) found only small to moderate increases in FL, with long length eccentric isokinetics resulting in the greatest adaptive response. This is despite similar weekly volumes being applied within the intervention, however, Guex and colleagues (2016) intervention lasted half the duration of the Timmins et al, (2016) intervention (3 weeks vs. 6 weeks); which could explain why the post-training results did not reach the same magnitude of change with the study by Guex and colleagues (Guex, Degache, et al., 2016).

A further 14 studies have examined maximal-supramaximal loading strategies including: bilateral to unilateral leg curl, NHE, unilateral flywheel and the glider. From these results, no change to very large increases were observed in FL for all studies (g = 0.00 – 5.61). Despite a general trend of increased FL, there are studies that only achieve trivial response, potentially not achieving a minimal meaningful change (Timmins, et al., 2015). As Seymore, Domire, De Vita, Rider, & Kulas (2017), and Lacome et al. (2019) found that intervention groups only achieved null to small changes in FL (g = 0.00-0.33), despite similar methodologies to previous studies. The large spread of observed magnitudes in FL change and high number of non-responsive groups could have a number of potential explanations, including, the prescription of the exercise with lower volumes of work performed potentially being preferential is achieving positive adaptations (Presland, et al., 2018). Seven of the ten studies also reported PA changes; however, the adaptive response was inconsistent between studies (Alonso-Fernandez, Docampo-Blanco, & Martinez-Fernandez, 2018; Duhig, et al., 2019; Guex, Degache, et al., 2016; Pollard, et al., 2019; Potier, et al., 2009; Presland, et al., 2018; Ribeiro-Alvares, et al., 2018; Seymore, et al., 2017; Timmins, Ruddy, et al., 2016), with both increases and decreases in PA observed. This finding potentially highlights that the overall adaptation of an increased FL, could be achieved via two distinct adaptations decreased PA or increased MT, although the latter is not commonly reported by researchers (Bourne, Duhig, et al., 2017). Additionally, athlete history or current preparedness could also be influencing the observed changes, with greater eccentric intensities, favouring positive adaptations in FL (Pollard, et al., 2019), however, it must be expected that a high current level of preparedness would be required (i.e. high levels of eccentric hamstring strength e.g. >400 N), before increasing eccentric intensity with the addition of load.

Elite youth male footballers, participated within study by Lacome, et al. (2019) this could indicate that they were highly accustomed to performing the NHE at bodyweight, potentially sub-maximally, signifying the minimal adaptive response could be due to insufficient intensity, requiring additional load for optimal progressive overload. Furthermore, the intervention period was performed in-season, which has already shown to effectively reduce BF_{LH} FL (Timmins, et al., 2017). Although it should be noted that trivial-small increases in FL were observed by Lacome, et al. (2019), recommending that to potentially offset the reducing effect from match and training demands, an increase in intensity and a decreased volume could be optimal. Similarly, the elite Portuguese soccer players who performed NHE intervention within Mendiguchia, et al. (2020) study could have also benefitted from an increased NHE training intensity or overload, which may have provided a larger positive magnitude in BF_{LH} FL than sprinting which was observed by the authors. Furthermore, the addition of acute augmented feedback (Chalker, et al., 2018), could increase athlete's motivation post-sport specific training, potentially playing a role to increase the observed positive adaptations within these studies (Pollard, et al., 2019).

Sprinting has been previously suggested to be a vaccine to HSI occurrence (Butler, 2019; Edouard et al., 2019), although to date only a single sprint training intervention study has been performed that has observed architectural changes in the BF_{LH} (Mendiguchia, et al., 2020). Elite Portuguese soccer players, performed two separate sprint training sessions per week for seven-weeks (force-velocity emphasis, gastrocnemius and an acceleration emphasis) (Mendiguchia, et al., 2020). The authors observed a moderate increase in BF_{LH} FL within the sprint training group (Mendiguchia, et al., 2020)**.** An earlier study by Freeman et al.(2019) demonstrated an increase in eccentric hamstring strength from a sprint intervention, which could be explained by the architectural adaptations seen by Mendiguchia, et al. (2020), although changes in strength were not observed. The extensive sprint training intervention used by Mendiguchia, et al. (2020) utilised both high velocity running along with the inclusion of higher force, slower velocity-based exercises, such as heavy sled towing (70% body weight) or bounding, which would consist primarily of concentric hip extension or lower level eccentric actions. The inclusion of these exercises, although very common in practice, could have had a negated some of the potential benefits of the high velocity running as concentric dominant tasks can lead to reductions in BF_{LH} FL (Bourne, et al., 2018; Duhig, et al., 2019; Timmins, Ruddy, et al., 2016). Additionally, bounding had a negative effect on HSI occurrence within soccer players (Van De Hoef, et al., 2019) – although BF_{LH} architectural properties and eccentric strength (i.e. modifiable risk factors) were not assessed at any time point by the researchers (Van De Hoef, et al., 2019). This study nevertheless provides novel evidence of a sprint intervention that could be implemented in elite sport for HSI risk mitigation (i.e. increased BF_{LH} FL), importantly, it did not encounter problems such as compliance, which was reported within the NHE group where participants were removed from the study for having <80% compliance (Mendiguchia, et al., 2020).

2.6.3.2 Alternative intervention options

A series of alternative methods that have also been utilised across the studies observing changes in BF_{LH} muscle architecture (5/12 studies), including concentric-only exercise (leg curl), concentric-eccentric task (single leg 45° hip extension), quasi-isometric exercise (Razor curl) and dynamic task (sprint acceleration) interventions. Concentric bias interventions can be separated into two distinct categories, knee vs hip dominant, or short vs long muscle length. Timmins et al, (2016) and Duhig et al, (2019) both utilised knee dominant concentric-only exercises, finding moderate to large decreases in FL. In contrast, Bourne et al., (2017) utilised a concentric-eccentric hip dominant exercise (single leg 45° hip extension) resulting in a large increase in FL, similar in magnitude to an identical volume NHE intervention (Bourne, Duhig, et al., 2017). The altered adaptive response from the concentric-eccentric hip dominant exercise (single leg 45° hip extension) could be due to the effect of training at a long muscle length (Bourne, Duhig, et al., 2017), with an increased working range. Whereas knee flexion tasks are typically performed at short to moderate muscle lengths. Furthermore, the positive FL adaptations observed by Bourne et al, (2017) could be the result of a different adaptive mechanism to the NHE. Specifically, the single leg 45° hip extension resulted in a significant increase in BF muscle volume, in comparison to the NHE (Bourne, Duhig, et al., 2017), leading to an increased FL via widening the distance between aponeuroses. However, changes in PA were not reported with the study to permit comparison to the eccentric adaptations in PA (Bourne, Duhig, et al., 2017).

Pollard and colleagues (2019) observed the effect of a quasi-isometric exercise (i.e. razor curl) on BF_{LH} muscle architecture. The authors found a trivial decrease in BF_{LH} FL, which may suggest that it is not an overly effective exercise – specifically with the aim of HSI risk mitigation (Pollard, et al., 2019). Although, despite not being an aim of the current chapter, the authors did report increases in eccentric hamstring strength, which may suggest that structural adaptations could still be occurring, whilst not altering FL especially at mid-muscle belly where US imaging is acquired (Pollard, et al., 2019). When considering a quasi-isometric action incorporates simultaneous movement across multiple joints, resulting in a constant muscle length (Pollard, et al., 2019; van den Tillaar, Solheim, & Bencke, 2017). Potentially signifying those structural adaptations could be occurring muscle at proximal and distal portions – although imaging at proximal and distal portions of the BF_{LH} is not common practice.

2.6.3.3 Intervention prescriptions – Duration, Intensity and Volume

Across the 16 studies, intervention durations ranged from 3 – 12 weeks, although a cross-over study design was employed by Lacome, et al. (2019), using two six-week training blocks. There was a non-significant, trivial-small effect observed with exercise duration, although the greatest magnitudes are observed within the longer durations (>6 weeks) even in closely matched studies such as the concentric training groups within the studies by Guex, Degache, et al. (2016) and Timmins et al, (2016) (Figure 2-9).

The shortest duration, of just 3-weeks, resulted in increases in BF_{LH} FL using an isokinetic eccentric training stimulus (Guex, Degache, et al., 2016). While a similar intervention performed by Timmins et al, (2016), which lasted for twice this duration (6-weeks) resulted in an increase in FL of more than twice the magnitude observed by Guex et al, (Guex, Degache, et al., 2016), highlighting that the duration of training is an important consideration. In contrast for the NHE, intervention duration has a limited influence upon increases in FL, with 4 – 12-week interventions resulting in similar increases in FL (Bourne, Duhig, et al., 2017; Ribeiro-Alvares, et al., 2018). However, on completion of the training interventions and cessation of the NHE, a period of just two-weeks has been identified as minimal time required for adaptations to return to pre-training BF_{LH} FL (Presland, et al., 2018). Therefore, consistent application of the NHE, is crucial in achieving and maintaining FL adaptations, important within team-based sports where increases in FL have a protective effect against HSIs, although the detraining effect of other modalities such as concentric-eccentric based tasks and sprinting have not been observed within the literature and could have a greater retention effect.

Typically, researchers that have used NHE within interventions, have looked to increase training volume or time under tension (i.e., NHE ability) as a form of progressive overload, without considering eccentric intensity (Petersen, et al., 2011; van der Horst, et al., 2015). However, for an exercise where increases in force generating ability is a desirable outcome, potentially achieved by architectural adaptations to the BF_{LH}, an increasing in working intensity could be considered a more effective method (Pollard, et al., 2019). Pollard et al, (2019) observed the effect of eccentric intensity of the NHE, finding that when performing the NHE an increased eccentric intensity (i.e. additional load), there was a greater change in BF_{LH} FL, in comparison to body weight alone. However, it must be presumed that these athletes were at an adequate level of strength, with a history of performing the NHE prior to commencing the intervention, whereby they could reach an appropriate degree of knee extension prior to falling with this additional load (Pollard, et al., 2019). This is the first intervention that has observed the effect of intensity when performing the NHE, while previous interventions have described when the NHE should be progressed (e.g., when the participant can control the exercise to the last 10 – 20° ROM) (Bourne, Duhig, et al., 2017; Duhig, et al., 2019). Despite this, studies that have limited the performance of NHE to bodyweight alone, have seen similar increases in FL, this implies that bodyweight alone is an adequate initial stimulus to achieve a positive response. However, there will come a point, as with any training stimulus, where the adaptations will plateau – requiring an increased intensity to progress further, with appropriate and progressive overload. This could explain the minimal changes observed by Lacome, et al. (2019) following the cross-over, where a change in training volume, did not provide an adequate stimulus to promote positive adaptations to BF_{LH} FL.

With regards to intensity of concentric resistance training – current studies have used between 60 – 83% of one repetition maximum (1RM) load (Bourne, Duhig, et al., 2017; Duhig, et al., 2019), for both a knee dominant and hip dominant tasks (Bourne, Duhig, et al., 2017; Duhig, et al., 2019). Bourne et al, (2017) demonstrated that training using a hip dominant exercise (single leg 45° hip extension), resulted in a significant meaningful increase in BF_{LH} FL – which contrasts that of knee dominant tasks (Duhig, et al., 2019; Timmins, Ruddy, et al., 2016). As previously mentioned, this could be related to the hip dominant tasks being performed at moderate to long muscle lengths. However, the effect of absolute loading could also play an important role in the observed adaptive response (i.e. positive change FL), as hip dominant tasks permit a greater absolute load capacity than any knee dominant tasks, having large contributions from supporting musculature such as the glutes and erector spinae (Andersen et al., 2018; Bourne, Williams, et al., 2017; Contreras, Vigotsky, Schoenfeld, Beardsley, & Cronin, 2015, 2016b; Jeon, Hwang, Jung, & Kwon, 2016; Korak, Paquette, Fuller, Caputo, & Coons, 2018; McCurdy, Walker, & Yuen, 2018; A. D. Vigotsky, Harper, Ryan, & Contreras, 2015).

The frequency and volume of training are also key considerations, across all intervention's frequency ranges from 1-3 x/week, Lacome, et al. (2019) and Medeiros et al. (2020) have employed a 1 x/week frequency – this could have been by design or by the constraints of their sporting environment (elite youth soccer). Nevertheless, Lacome, et al. (2019) had the lowest adaptive response for any NHE intervention, contrastingly, Medeiros, et al. (2020) found a small effect positive effect, similar in magnitude to performing the intervention twice per week. Although as discussed other factors could explain the minimal response, such as intensity. However, this is an important consideration, as a minimum required frequency of ≥2 x/week could be essential in achieving greater adaptive responses with the NHE. With regards to training volume a large range has been used across the studies, with average weekly doses across all resistance training interventions range from 10 – 97 repetitions, whist volume per sessions ranges from 8 – 100 repetitions. A low volume approach as prescribed by Presland et al. (2018), had the greatest positive effect of BF_{LH} FL in comparison to high volume equivalent, although this could be an effect of an initial higher, matched volume control period where a large rebound or supercompensation could have occurred. With regards to an optimal training volume, it is difficult to come to a conclusion with the present literature; as various volumes, frequencies and intensities have been utilised. However, as HSIs are a frequent problem within team-based sports with congested fixture and training schedules, a dose which achieves the positive adaptations to BF_{LH} muscle architecture and eccentric hamstring strength – while minimising muscle soreness and fatigue would be considered optimal. For sprint-based interventions, a recommended volume is potentially even more complex, as the extensive approach used by Mendiguchia, et al. (2020) was effective. The question around sprinting is do we need to substitute sets x reps for a sufficient distance over a set number of repetitions, however at this early stage in the research an answer is currently unclear. If using sprint training to mitigate the risk of HSI, the aim should be to prepare athletes to all potential sprinting demands, so this would include acceleration and high velocity-based running tasks, with short and long distances with different approach set ups (e.g., walk-ins, flying starts, sprint-float-sprint). Additionally, the potential of using sport specific set ups, including ball pick-ups and hip flexed running again with the goal of maximising athlete preparedness also should be investigated with regards to the modifiable risk factors of HSI. However, large variations in weekly training and match sprint volumes can significantly influence HSI incidence (Malone, et al., 2018), therefore, it is crucial not to "overcook" players, but to be adaptable to training volumes with changing situations in practise. Additionally, although Freeman et al. (2019) did not observe muscle architecture, the intervention was far less extensive with lower volumes of only maximal sprinting utilised.

2.6.3.4 Ultrasound assessment method considerations

The methods that have been employed within the studies selected for this review, to measure and estimate BF_{LH} FL, all studies used probe lengths of < 5 cm (4.1 – 4.7 cm). This is an important methodological detail as smaller probes result in an increased degree of estimation, with the reduced field of view (FOV), potentially increasing measurement error (Franchi, Fitze, et al., 2019; Franchi, et al., 2018). Despite the short probe lengths utilised, there were two imaging techniques: single image estimation and panoramic imaging. Both imaging methods have varying degrees of measurement error (Franchi, Fitze, et al., 2019), despite panoramic imaging potentially being more accurate than single image estimation it does require high levels of sonographer expertise, while the additional analyses required for panoramic imaging can be time consuming indicating it may not be best suited for practitioners who may be limited by such. Furthermore, from single image estimation, two methods of have been utilised, FL estimation equation 2-1 and linear extrapolation. Equation 2-1 requires the greatest degree of estimation and despite being reliable is not necessarily the most accurate (De Oliveira, Carneiro, & De Oliveira, 2016; Franchi, Fitze, et al., 2019; Freitas, et al., 2018; Pimenta, et al., 2018), as it requires the estimation of the complete fascicle. Despite linear extrapolation being a more subjective method of estimation, it does however limit the degree of estimation as you are only estimating the un-viewable portion (Franchi, Fitze, et al., 2019). Although with both of these methods, there is the assumption of no fascicle curvature which is where panoramic US or a probe with a greater FOV could be more effective.

2.6.4 Conclusions

To my knowledge, this is the first review of the literature across training interventions on BF_{LH} architectural adaptations as a result of imposed training interventions. In order to mitigate HSI risk, practitioners working within sports that have elements of high velocity running (i.e. team sports, track sports, etc), should aim to increase BF_{LH} FL within athletes to optimise the contractile components and operating characteristics of the muscle (force-length and force velocity relationships) (Bourne, et al., 2018; Timmins, Shield, Williams, Lorenzen, et al., 2016). Across the interventions utilised, eccentric, hip dominant conventional concentric-eccentric training (e.g. 45° hip extension) and sprinting methods appear to be the most effective training methods to increase BF_{LH} FL. Practitioners could look to increase BF_{LH} FL by adopting a variety of methods within a mixed modal approach, although a mixed modal approach has never been utilised within the research, with only single mode interventions utilised – which is dissimilar to practice where a single methodology would not commonly be implemented. Therefore, research is required to observe the effect of mixed method approach (i.e., NHE, hip dominant conventional concentric exercises and/or sprinting).

2.7 Systematic review and meta-analysis of the Effect of Hamstring Training Interventions on Eccentric Hamstring Strength

Alongside BF_{LH} FL, the eccentric force producing capacity of the hamstrings is also key in reducing HSI risk and occurrence (Bourne, et al., 2018; Opar, et al., 2015; Timmins, Bourne, et al., 2016). To date, numerous intervention modalities have been investigated, however, the effect of each intervention on eccentric hamstring strength have never been compared. Results of a recent systematic review and meta-analysis identified and compared NHE based interventions on both force and BF_{LH} FL adaptations, concluding that the NHE results in large positive effects on eccentric force or torque (Cuthbert, et al., 2019). Although the same result was not found for relative eccentric torque, with only trivial to small adaptations found (Cuthbert, et al., 2019). Nevertheless, comparisons between different resistance training modalities, such as the NHE, sprinting and isokinetics have never been made. Therefore, the aim of this review is to systematically review the literature across resistance-based training interventions which measured the changes on eccentric hamstring strength on athletic individuals.

2.7.1 Methods

2.7.1.1 Literature search

A systematic, computerised search of the literature in PubMed, SPORTDiscus, MEDLINE, Scopus and Web of science was conducted, with controlled vocabulary and key words related to hamstring injury prevention programmes and hamstring injury. Our search timeframe was from inception to March 2020. Key words (Table 2-7) were chosen in accordance with the aims of the research. Search terms were combined by Boolean logic (AND [between categories], OR [within categories]). We also extended the search spectrum to "related articles" and the bibliographies of all retrieved studies.

Table 2-7 Summary of keyword grouping employed during database searches.

Anatomy	Physical quality	Imposed demand	Methodology
Hamstring	Strength	Resistance training	Isokinetic
Posterior thigh	Eccentric	Eccentric	Nordbord
Bicep femoris		Concentric	Strain gauges
		training	

2.7.1.2 Selection criteria

The following inclusion criteria were used to select articles focused on the effect of training and HSI on eccentric hamstring strength:

1. Full-text, research articles exploring, and analysing adaptations eccentric hamstring strength were selected. As such, case studies, review articles, and abstracts were excluded.
2. Research articles must report changes in eccentric hamstring strength.

2.7.1.3 Quality assessment

Study quality was evaluated by a standard procedure (Table 2-8). Each study was read and ranked from 0 to 5, with the larger number indicating better quality. For each question, a 1 was awarded if the study met the standard. If insufficient description or data were provided to analyse a specific question, a 0 was awarded. The score was the tallied for each question, with the highest score possible equalling 5 out of 5.

Table 2-8 Quality assessment of eccentric hamstring strength intervention studies.

			Score
A	**Sample description:**		0 or 1
	+	Properties of the subjects (age, weight, height, sex)	
	+	Definition of the population (well-trained, recreationally trained, untrained)	
B	**Intervention:**		0 or 1
	+	Defined and supervised training programme (exercise performance, coaching, progressions)	
	+	Defined volume and frequency	
C	**Methods employed for eccentric hamstring strength assessment:**		0 or 1
	+	Defined methodology (assessment type, joint angles, velocities etc.)	
D	**Data analysis**		0 or 1
	+	Defined analysis processes (software, units)	
E	**Results detailed:**		0 or 1
	+	Measure of central tendency and variation from the average	

2.7.1.4 Statistical analyses

To assess the magnitude of each training stimulus, where possible Hedge's *g* effect sizes (ES) were calculated from the mean and standard deviation and sample size. Where the mean and standard deviation were not reported by the authors, with only magnitude-based differences described, typically Cohen's *d* ES, conversions to Hedge's *g* ES were made to allow comparison between studies and training modalities (Equation 2-1) (Lakens, 2013).

$$Hedge'sg = Cohen's\ d \times (1 - \left(\frac{3}{4 \times (n-1) - 1}\right))$$

Equation 2-2 Effect size conversion from Cohen's d to Hedge's g.

Alongside the magnitude of differences, the mean difference was also reported for each study. The scale for interpretation of ES was proposed by Hopkins (2010) as follows; trivial (≤ 0.20), small (0.21–0.59), moderate (0.60–1.19), large (1.20–1.99), or very large (≥2.00).

DerSimonian and Laird (1986) random effects models was used to observe the overall effect using the Z statistic. Consistency of effects was quantified using a test for heterogeneity (I^2) outlined by Higgins, et al. (2003) whereby a scale of low (<25%), moderate (25-75%) and high

(>75%) I^2 values were used for the interpretation of consistency. The duration of study intervention was used as a moderator within the DerSimonian and Laird (1986) random effects model, to observe the effect of study duration on the magnitude of adaptations.

2.7.2 Results

2.7.2.1 Search Results

A flow diagram of the literature search and the final selection is shown in Figure 2-11. According to the above-defined inclusion criteria, 24 independent studies were identified in which changes in eccentric hamstring strength from hamstring strength training interventions were reported.

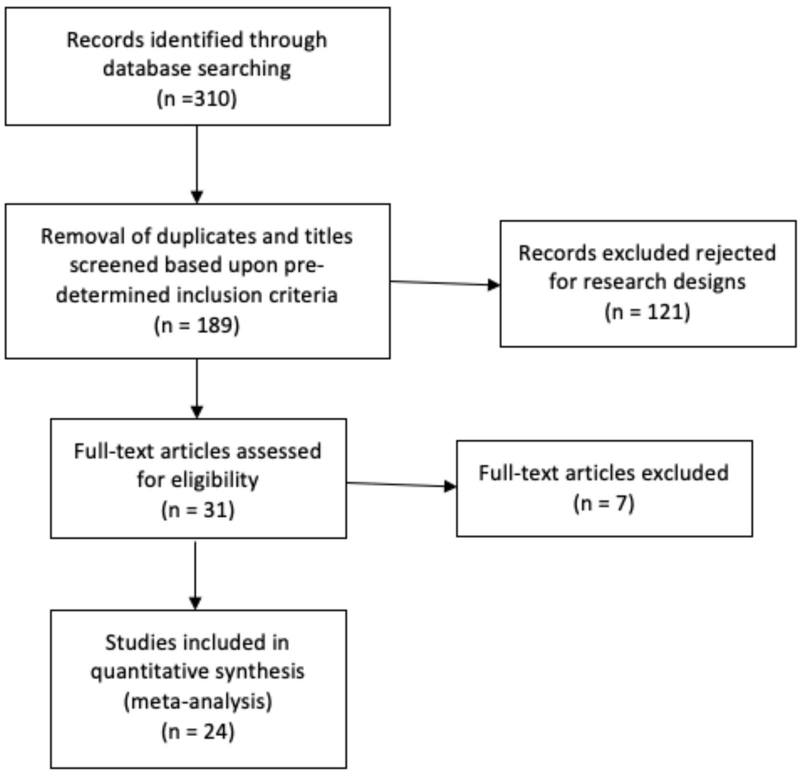


Figure 2-11 Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) flow diagram for study inclusion.

2.7.2.2 Study description

The characteristics of the studies are presented in Table 2-9.

Table 2-9 Individual training intervention study characteristics, descriptive, magnitude of change and quality assessment for eccentric strength

Study		*n*	Participant's characteristics				Intervention prescription						Eccentric strength assessment	Study quality	Outcome measures			Hedge's *g* (95% CI)
			Study population description	Age (years)	Height (cm)	Mass (kg)	Length (Weeks)	Modality	Average weekly dose	Frequency per week	Volume per session range	Intensity			Unit	PRE (SD)	POST (SD)	
2003	**Askling**	15	Professional Swedish footballers	24.0 ± 2.6	182.0 ± 6.0	78.0 ± 5.0	10	Flywheel	32	1	32	Deceleration velocity	IKD@60	4	n/m	148 (24)	176 (22)	1.15 (-0.01 - 2.18)
2004	**Mjølsnes**	11	Student competitive soccer players	**NR**	**NR**	**NR**	10	NHE	91.6	1-3	10-30	No progression	IKD@60	4	n/m	240 (12)	267 (13)	1.99 (0.47 - 3.46)
2012	**Iga**	10	Professional footballers	23.4 ± 3.3	177.0 ± 7.0	78.0 ± 8.2	4	NHE	40	1-3	10-24	No progression	IKD@60	5	n/m	115 (42)	132 (43)	0.37 (-0.89 - 1.61)
													IKD@120			121 (45)	134 (42)	0.27 (-0.98 - 1.51)
													IKD@240			121 (43)	130 (42)	0.19 (-1.06 - 1.43)
													IKD@60			99 (30)	119 (37)	0.54 (-0.74 - 1.79)
													IKD@120			105 (32)	119 (33)	0.39 (-0.88 - 1.63)
													IKD@240			102 (34)	122 (32)	0.55 (-0.38 - 1.88)
2016	**Delahunt**	15	Rec-active	22 ± 1.38	182 ± 5.0	78.37 ± 8.54	6	NHE	56.7	1-3	10-30	No progression	IKD@120	4	n/m	177.4 (27.9)	204.4 (30.5)	0.87 (-0.21 - 1.92)
													IKD@120		n/m/kg	2.25 (0.32)	2.59 (0.36	0.94 (-0.15 - 2.00)
	Guex	11	Rec-active	28.4 ± 4.5	170.7 ± 5.9	64.0 ± 12.7	3	Long length eccentric IKD	64	2-3	24	No progression	IKD@60	4	n/m	52.1 (23.9)	60.2 (45.3)	0.20 (-0.99 - 1.38)
		11		27.3 ± 3.9	173.5 ± 10.8	66.0 ± 13.6		Short length eccentric IKD	64		24		IKD@60			59.4 (22.9)	65.5 (21)	0.26 (-0.93 - 1.44)
	Guex	10	National sprinters	20.7 ± 2.2	175.9 ± 5.8	64.9 ± 4.2	6	IKD knee flexion and hip flexion	80	1-2	32-60	No progression	IKD@30	4	n/m/kg	1.8 (0.31)	2.13 (0.28)	1.02 (-0.34 - 2.33)
		10						IKD knee flexion and hip flexion					IKD@120			1.89 (0.32)	2.26 (0.24)	1.20 (-0.20 - 2.54)

2017	**Bourne**	10	Rec-active	22.0 ± 3.6	181.4 ± 7.0	80.8 ± 11.1	10	NHE	72.6	2	12-50	2.5kg Incremental loading	NB	5	N	**NR**	**NR**	2.16 (0.50 - 3.74)
		10						45deg hip extension				60-80% RM	NB			**NR**	**NR**	1.15 (-0.24 - 2.48)
	Seymore	10	Rec-active	18.3 ± 0.5	170.0 ± 10.0	71.3 ± 15.9	6	NHE	62.6	1-3	10-30	No progression	IKD@60	5	n/m/kg	1.55 (0.57)	1.73 (0.8)	0.24 (-1.01 - 1.48)
2018	**Alt**	16	Regional-national sprinters	21.6 ± 2.5	181.4 ± 7.1	75.7 ± 9.8	4	Mix of harness assisted and unassisted NHE	27	3	9	Assisted too Unassisted	IKD@15	5	n/m/kg	2.04 (0.24)	2.17 (0.26)	0.49 (-0.51 - 1.48)
													IKD@60			2.02 (0.29)	2.18 (0.32)	0.50 (-0.51 - 1.48)
													IKD@150			2.02 (0.28)	2.18 (0.31)	0.51 (-0.50 - 1.50)
	Ishoi	11	Amateur Danish footballers	19.1 ± 1.8	180.7 ± 7.3	76.2 ± 11.9	10	NHE	91.6	1-3	10-30	No progression	NB	5	N	321.5 (50.7)	383.2 (68.6)	0.98 (-0.31 - 2.22)
	Matthews	11	Collegiate soccer players	21.8 ± 2.8	179.4 ± 6.5	79.3 ± 11.8	4	NHE	40	2	20	No progression	IKD@60	5	n/m	151.55 (13.72)	165 (25.58)	0.69 (-0.55 - 1.90)
																118.64 (15.53)	162.82 (30.18)	1.62 (0.20 - 2.98)
		9		23.2 ± 3.8	184.8 ± 8.4	82.5 ± 8.8		Band assisted NHE	120	2	60	No progression				99.78 (10.1)	132.78 (25.29)	1.58 (0.00 - (3.08)
																122.22 (4.68)	137.56 (21.93)	0.89 (-0.52 - 2.25)
	Presland	10	Rec-active	22.3 ± 2.8	179.1 ± 7.7	75.1 ± 8.8	4	NHE	73.3	2	24-50	2.5kg Incremental loading	NB	5	N	410 (53)	536 (104)	1.40 (-0.04 - 2.78)
																403 (49)	501 (88)	1.26 (-0.15 - 2.61)
		10							21.3	1	8-48					432 (55)	587 (74)	2.17 (0.51 - 3.75)
																423 (61)	553 (86)	1.59 (0.10 - 3.01)
	Siddle	8	Amateur team sport athletes	20.47 ± 1.32	179.81 ± 7.45	75.54 ± 7.1	6	NHE	36.6	1-2	10-27	No progression	IKD@180	5	n/m	133.13 (18.34)	164.94 (21.29)	1.40 (-0.23 - 2.94)
	Riberiro-Alvares	10	Moderately active students	23.7 ± 3.3	165.1 ± 9.0	59.1 ± 12.8	4	NHE	46.5	2	18-30	No progression	IKD@60	5	n/m	110.9 (21.2)	126.9 (33.4)	0.52 (-0.76 - 1.77)
2019	**Duhig**	15	Rec-active	22.8 ± 4.1	180.1 ± 6.4	85.2 ± 14.6	5	NHE	39.6	1-2	12-30	5kg Incremental loading	NB	5	n/m/kg	3.69 (0.53)	4.55 (0.55)	1.55 (0.36 - 2.70)
		15						Concentric unilateral knee flexion				6-7RM				3.88 (0.47)	4.4 (0.62)	0.92 (-0.17 - 1.98)

2019	**Freeman**	14	Adolescent team sport athletes	16.2 ± 1.3	175.0 ± 10.0	68.5 ± 12.1	4	NHE	25.2	2	3-18	No progression	NB	5	N	329 (75.8)	361.1 (89.4)	0.36 (-0.70 - 1.41)
		14						Sprinting	498 m	2	80-400 m	No progression				315.4 (44.8)	335.1 (74.7)	0.25 (-0.81 - 1.30)
	Suarez-Arrones	17	Professional Spanish footballers	18.8 ± 0.8	176.8 ± 6.9	71.3 ± 5.7	15	NHE	37.9	1-2	10-30	No progression	NB	4	n/m/kg	9.52 (0.88)	9.78 (1.01)	0.26 (-0.70 - 1.21)
		17													N	692.5 (90.9)	702.6 (87.9)	0.11 (-0.84 - 1.06)
		16					17		36.9	1-2	10-30	No progression			n/m/kg	8.46 (1.51)	9.64 (1.41)	0.77 (-0.26 - 1.78)
		16													N	570.5 (106.5)	660.0 (105.5)	0.80 (-0.24 - 1.81)
	Pollard	10	Rec-active	24.4 ± 4.0	181.0 ± 6.0	78.0 ± 11.0	4	Body weight NHE	21.3	1	8-48	No progression	NB	5	N	460 (112)	528 (87)	0.62 (-0.67 - 1.88)
		10						Weighted NHE				2.5kg Incremental loading				465 (96)	546 (78)	0.85 (-0.48 - 2.13)
		10						Razor curl								441 (75)	506 (82)	0.76 (-0.55 - 2.03)
2020	**Presland**	10	Rec-active	27.8 ± 5.3	178.4 ± 7.7	80.0 ± 10.7	6	Eccentrically loaded Unilateral Flywheel	65.3	2	48-96	0.1 kg.m	NB	5	N	440 (110)	473 (86)	0.31 (-0.95 - 1.55)
								Conventional Unilateral Flywheel (contralateral limb)				0.05 kg.m				435 (93)	478 (92)	0.48 (-0.80 - 1.73)
		10						Conventional Unilateral Flywheel				0.05 kg.m				499 (75)	541 (85)	0.43 (-0.84 - 1.67)
	Delextrat	9	Collegiate hockey players	19.7 ± 1.4	168.4 ± 4.4	66.2 ± 7.2	6	NHE	61	3	12-30	No progression	IKD@120	5	n/m - D	55.4 (15.8)	62.2 (19.4)	0.35 (-0.91 - 1.50)
															n/m - ND	43.2 (18.4)	56.1 (16.9)	0.67 (-0.63 - 1.93)
		8		19.5 ± 1.0	168.1 ± 3.4	66.7 ± 4.5	6	Eccentric leg curl	61	3	12-30	No progression			n/m - D	60.9 (18.8)	65.2(16.4)	0.22 (-1.03 - 1.46)
															n/m - ND	50.9 (13.3)	66.4 (16.9)	0.93 (-0.41 - 2.22)

2020	**Drury**	8	Youth soccer players	11.0 ± 0.9	144.2 ± 4.4	37.7 ± 2.8	6	NHE	27	1-2	10-24	No progression	NB	5	N/kg	4.27 (0.88)	4.95 (0.76)	0.74 (-0.73 - 2.16)
		16		14.0 ± 1.1	173.2 ± 7.0	61.8 ± 6.3	6								N/kg	4.69 (0.85)	5.17 (0.95)	0.50 (-0.50 - 1.50)
	Marušič	18	Healthy individuals	23.4 ± 3.3	177.0 ± 7.0	78.0 ± 8.2	6	LL eccentrics (Modified NHE and glider)	37.3	1	20-48	5-20kg Incremental loading	IKD@60	5	N/m /kg	1.79 (0.55)	2.09 (0.52)	0.54 (-0.41 - 1.47)
													IKD@180			1.70 (0.51)	1.90 (0.41)	0.41 (-0.53 - 1.34)
													NB			2.29 (0.76)	3.04 (0.55)	1.08 (0.07 - 2.06)
	Medeiros	15	U20 and U23 professional footballers	18.80 ± 1.74	182.0 ± 8.0	78.80 ± 8.39	8	NHE	28.5 - 30.5	1	12 - 40	No progression	IKD@60	5	N/m	216.29 (31.35)	217.86 (38.24)	0.04 (-0.97 - 1.05)
		17		18.47 ± 1.07	179.0 ± 10.0	75.53 ± 10.44	8		57-61	2					N/m	197.48 (40.81)	216.39 (34.17)	0.48 (-0.49 - 1.44)
Published ahead of print	**Severo-Silveira**	11	Competitive rugby players	27.20 ± 3.26	175.0 ± 5.0	90.10 ± 14.30	8	Constant volume NHE	34.5	2	12-18	No progression	IKD@60	4	N/m	204.58 (43.42)	207.01 (41.67)	0.05 (-1.14 - 1.34)
		10		25.20 ± 3.34	176.0 ± 8.0	88.60 ± 12.80	8	Incremental volume NHE	57-61						N/m	211.17 (31.81)	225.64 (43.29)	0.35 (-0.86 - 1.52)

Trivial - <0.20	Small - 0.20–0.59	Moderate - 0.60–1.19	Large - 1.20–1.99	Very Large - >2.00

NR = Not reported within the study, § = Crossover study design, SL = Short muscle length, LL = Long muscle length, NHE = Nordic hamstring exercise, SLDL = Stiff leg deadlift, RM = One repetition maximum, BW = Bodyweight, IKD@ = isokinetic at velocity, NB = Nordbord

2.7.2.3 Magnitude (Hedge's g) of change in eccentric hamstring strength

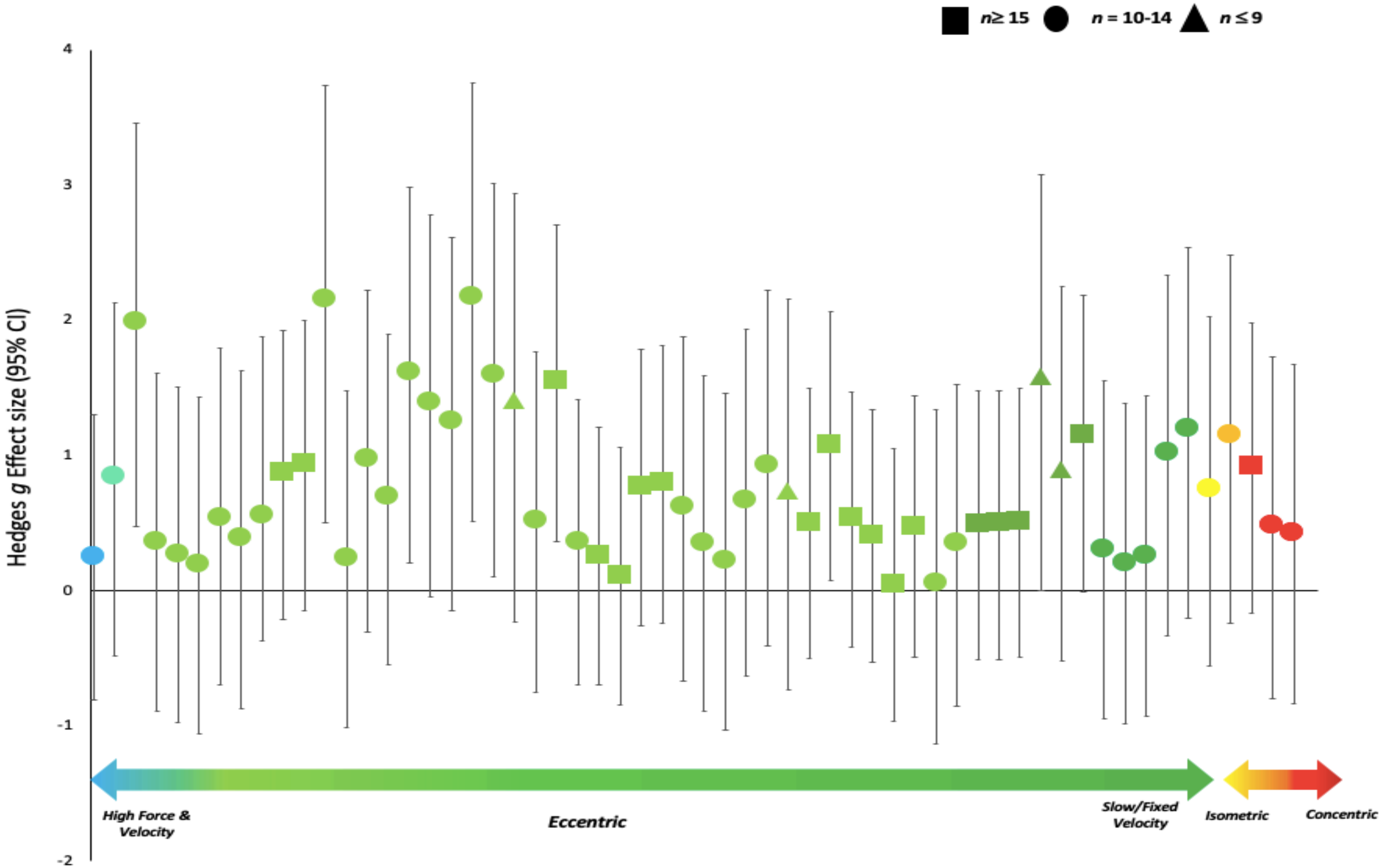


Figure 2-12 Magnitude of effects across training interventions, distinguishing primary muscle actions stimulus provided

All interventions were found be to positively effective, with increases across all modalities. Although, a greater likely adaptation was found for eccentric modalities, where a greater proportion of the studies 95% CIs did not cross the zero line (Figure 2-12). The overall pooled estimate (Hedge's *g*) from the main effects analysis was 0.72 (95% CI 0.44 to 1.00). The test for overall effect favoured the intervention treatments (Z = 5.04, $p < 0.001$). There was no consistency observed between the studies (I^2= 0%, p = 1.00). When using study duration as a moderator within the DerSimonian and Laird (1986) random effects model, there was a no effect (0.01 (95% CI -0.15 to 0.17)), with the test for overall effect (Z = 0.10, p = 0.922). When using modality as a moderator within the DerSimonian and Laird (1986) random effects model, there was a small effect (0.48 (95% CI -1.55 to 2.51)) favouring eccentric modalities (eccentric and sprinting), with the test for overall effect (Z = 0.46, p = 0.646).

2.7.3 Discussion

The first notable observation from the present systematic review, is that regardless of the intervention modality utilised (eccentric, isometric or concentric), there were positive increases in eccentric hamstring strength; irrespective of assessment method (isokinetic or Nordbord) or unit of measure (absolute or relative to body mass). This highlights the first key finding from this systematic review, that training increases eccentric hamstring strength – potentially reducing the risk of sustaining a future HSI (Bourne, Duhig, et al., 2017). Although, similar to BF_{LH} FL (Chapter 2.6), there are number of influencing factors to a training modalities effectiveness (i.e., increased magnitude of change).

2.7.3.1 Eccentric Strength

It should come as no surprise that the training modality that resulted in the greatest magnitude of increase in eccentric hamstring strength, was eccentric modalities. However, within the literature there are a number of eccentric modalities that have been utilised, ranging from slow, fixed velocity eccentrics (e.g., isokinetic training) to high velocity eccentric actions within the terminal swing phase of sprinting. To date, the most commonly implemented intervention within the literature utilises moderate-high volume NHE interventions, using bodyweight alone (Delahunt, McGroarty, De Vito, & Ditroilo, 2016; Freeman, et al., 2019; Iga, Fruer, Deighan, Croix, & James, 2012; Ishoi, et al., 2018; Matthews et al., 2017; Mjølsnes, et al., 2004; Ribeiro-Alvares, et al., 2018; Seymore, et al., 2017; Siddle et al., 2018; Suarez-Arrones et al., 2019). However, similar to BF_{LH} FL, where Pollard et al, (2019) observed that performing the NHE with an increased eccentric intensity (i.e., additional load), resulted in a greater change in BF_{LH} FL and eccentric hamstring strength, in comparison to body weight alone. Furthermore, the studies which actively progressed the eccentric intensity with the addition of load also reported the greater magnitude ($g \geq 1.40$). Only two body weight alone NHE interventions have reported similar large magnitude increases in eccentric hamstring strength, the first Mjølsnes, et al. (2004) (coincidently, the first study to have implemented a NHE intervention), reported utilising extremely, high volumes of the NHE, with an average weekly dose of 91.6 repetitions per week. The second study, Matthews, et al. (2017) found a similar large magnitude increase in eccentric hamstring strength (g = 1.62), however this was only observed for the weaker limb, where the stronger limb only reported a moderate increase (Matthews, et al., 2017). Progressive overload refers to assigning training that is of a greater intensity than the athlete or individual is accustomed to (Sheppard and Triplett, 2016), therefore, progressive overload of the NHE would require a greater working intensity to continually see improvements (Pollard, et al., 2019; Sheppard and Triplett, 2016). This could be achieved within the first few weeks of performing the NHE, where after an initial exposure to the NHE, there would be a progression of expressed force as there would be an increase in the controlled range of motion. However, when individuals are capable of tolerating such loads (capable of controlling the last 10-20° ROM during the NHE), the addition of load in small increments would be essential to continually see improvements (Pollard, et al., 2019; Sheppard and Triplett, 2016). Additionally, this could be achieved by altering the muscle length of the NHE (flexing at the hip) or performing on an angled slope to increase the forces experienced at any given joint angle (Sarabon, Marusič, Marković, & Kozinc, 2019), although further research is required to determine if these are effective methods at providing overload to the NHE.

High velocity eccentric training (which have been proposed to occur during sprinting) is a novel avenue of research with regards to the modifiable risk factors of HSI (i.e. BF_{LH} FL and eccentric hamstring strength – despite a number of reviews and practitioners explaining why sprinting could be the answer earlier in 2016 (Butler, 2019; Edouard, et al., 2019; Edouard, Samozino, et al., 2016; G Moir, Brimmer, Snyder, Connaboy, & Lamont, 2018; Morin and Edouard, 2017; Oakley, Jennings, & Bishop, 2018), however, the research supporting this has never been followed up until recently. To date there are only two studies published, Mendiguchia, et al. (2020) who observed BF_{LH} FL changes and Freeman et al. (2019) who observed eccentric hamstring strength changes, following sprint training interventions. Freeman and colleagues (2019) demonstrated that a sprint training intervention does increase eccentric hamstring strength (g = 0.93), similar to the observations around BF_{LH} FL (g = 0.25) (Mendiguchia, et al., 2020). However, contrasting Mendiguchia, et al. (2020) who identified that a sprint training intervention is more effective at increasing BF_{LH} FL than the NHE. Whereas the sprint training intervention by Freeman, et al. (2019) resulted in a lower adaptive response when compared to the NHE. Although it should be noted that the sprint training group, initially started off considerably slower than the NHE training group at the top end (30-40 m time) (Freeman, et al., 2019). Therefore, being the "slower" athletes during the later phases of the sprint, could result in a lower intensity eccentric action (Heiderscheit, et al., 2005; Higashihara, et al., 2019; Kenneally-Dabrowski, Brown, Warmenhoven, et al., 2019; Nagano, et al., 2014; Schache, et al., 2012; Schache, Dorn, Wrigley, Brown, & Pandy, 2013), hence, it may not have provided the stimulus required for adaptation. The sprint training group athletes may have benefitted more from sprint technique modification, including increasing forward lean, and reducing upper body rotation, which could further enhance the benefits of sprint training by facilitating a greater application of horizontal force for improvements in the acceleration phases of running. Additionally, the duration of the study was low, only four weeks, therefore a longer duration may have also contributed to a greater magnitude of change (Freeman, et al., 2019).

Eccentric exercise such as the NHE are supramaximal in nature (i.e., where any muscle action cannot overcome the demand of the observed forces) – however, the systematic literature search identified two studies that have utilised sub-maximal eccentric exercises (i.e., assisted NHE) (Alt, Nodler, Severin, Knicker, & Struder, 2018; Buchheit, Simpson, Hader, & Lacome, 2019; Matthews, et al., 2017). Alt and colleagues (2018) used a harness assisted system, which was inter-mixed across the intervention with traditional NHE repetitions, they observed similar small increases (*g* = 0.49-0.51) in eccentric hamstring strength, highlighting that this type of intervention has benefits across both slow (15 deg/s) and fast (150 deg/s) isokinetic velocities (Alt, et al., 2018). Matthews, et al. (2017) used two intervention groups; traditional NHE and band assisted NHE, both groups had moderate increases in eccentric hamstring strength within the stronger limb, with large increases within the weaker limb (Matthews, et al., 2017). This firstly indicates, that although the NHE is a bilateral task it can have very specific unilateral adaptations and future studies both look to assess both limbs independently, in place of a mean change. It is also worth noting that the band assisted NHE training group were extremely weak pre-intervention (weak limb = 99.78$\pm$10.10 N·m , strong limb = 122.22$\pm$4.68 N·m), and even at the post-intervention did not reach the same absolute levels of eccentric strength of the traditional NHE training groups strong limb (Matthews, et al., 2017). Furthermore, the band assisted NHE training group performed triple the weekly volume than the traditional NHE training group (40 Vs 120 repetitions/week) (Matthews, et al., 2017). This could suggest that the identical magnitude in change observed were an effect of the increased training volume performed by the band assisted training group, however until a volume matched intervention study is performed, it is unknown. Although, due to the decreased working intensity of the band assisted NHE, it does permit an increased working volume without the associated DOMS or soreness and should be considered. Further methodological shortcomings should also be highlighted within the study by the Matthews, et al. (2017), specifically the utilisation of an elastic band – with potential for band fatigue. The methods also highlighted that the band was held by the investigator (Matthews, et al., 2017), which was in no way standardised negatively effecting the repetition-by-repetition assistance, further limiting the ability to perform study replication.

2.7.3.2 Alternative training modalities

Consistent with the previous systematic review observing BF_{LH} FL adaptations, eccentric modalities are not the only ones that have been applied within the literature, with quasi-isometric, traditional concentric-eccentric and concentric only modalities being identified within the literature (Bourne, Duhig, et al., 2017; Bourne, et al., 2018; Duhig, et al., 2019; Pollard, et al., 2019). As highlighted, regardless of the training modality there were increases in eccentric hamstring strength, traditional concentric-eccentric exercise (45° single leg hip extension) had the greatest effect on eccentric hamstring strength albeit a small increase (*g* = 1.15) (Bourne, Duhig, et al., 2017); interestingly this a hip dominant task that increased eccentric hamstring strength, assessed as the knee joint (Nordbord) (Bourne, Duhig, et al., 2017). Small increases in eccentric hamstring strength were also observed for concentric only knee flexion and a quasi-isometric exercise (razor curl) (*g* = 0.92 and 0.76, respectively) (Duhig, et al., 2019; Pollard, et al., 2019). These findings contrast that of the BF_{LH} FL adaptations, whereby a concentric only knee flexion and a quasi-isometric exercise (razor curl) resulted in reductions in BF_{LH} FL potentially elevating the risk of future HSIs (Duhig, et al., 2019; Pollard, et al., 2019). Despite increasing eccentric hamstring strength – this could signify that there is a trade-off in training between eccentric hamstring strength and BF_{LH} FL, when incorporating alternative modalities other than eccentric training. This is an important finding for practitioners, as multiple modalities are normally incorporated into resistance training programmes.

2.7.3.3 *Intervention prescriptions – Duration, Intensity and Volume*

As was observed with BF_{LH} FL, the duration of intervention studies had a large range from a minimum of three weeks, lasting up to 17 weeks (Suarez-Arrones, et al., 2019). Despite study duration having no effect, when used as a moderator within the DerSimonian and Laird (1986) random effects model, it could be that the large study variances (CIs) and large differences in assessment methods and training could explain why in its current format, study duration had no effect when used as a moderator.

It should be noted that similar to BF_{LH} FL interventions, Guex et al. (2016) remained the shortest duration intervention (three weeks) finding a small magnitude of adaptation to isokinetic eccentric strength. However, a similar eccentric isokinetic intervention study performed by the same research group (Guex, Lugrin, Borloz, & Millet, 2016), which was six weeks in duration found moderate and large increases in eccentric hamstring strength at both slow (30 deg/s) and fast (120 deg/s) isokinetic velocities, respectively. This highlights that intervention duration is a primary factor in training effectiveness when using isokinetic methods, for both modifiable risk factors of HSI (i.e., eccentric hamstring strength and BF_{LH} FL). However, when observing more applied methods (e.g., NHE), intervention duration had a limited influence upon increases observed within eccentric hamstring strength, with both trivial to very large increases across intervention durations (Table 2-9). A cessation period of just two-weeks has been identified as minimal time required for adaptations to return to pre-training BF_{LH} FL (Presland, et al., 2018), the same is not true with eccentric hamstring strength with a slightly longer retention period – although there were small decreases in peak eccentric force identified (Pollard, et al., 2019; Presland, et al., 2018; Siddle, et al., 2018). Despite not being as crucial, consistent application would still be advised to provide a continual improvement or at the very least, maintenance of eccentric strength capabilities.

For training intensity, NHE interventions typically increased training volume or ROM/time under tension (i.e., NHE ability), in place of eccentric intensity, via the addition of load (Pollard, et al., 2019). This makes minimal sense from a strength perspective (Sheppard and Triplett, 2016), where an increase in load would be considered desirable, if not essential in order to increase force producing capabilities. Again, Pollard et al, (2019) identified that the addition of load (i.e., increased eccentric intensity), resulted in a greater change in eccentric hamstring strength, in comparison to body weight alone. Previous interventions have described when the NHE should be progressed (e.g., when the participant can control the exercise to the last 10 – 20° ROM) (Bourne, Duhig, et al., 2017; Duhig, et al., 2019), with these studies resulting in large to very large increases in eccentric hamstring strength, even within extremely strong athletes (Peak eccentric force >500N) (Pollard, et al., 2019). Although, as with BF_{LH} FL, studies that have limited the performance of NHE to bodyweight alone increasing volume, ROM or time under tension, have seen similar increases in eccentric hamstring strength, highlighting that bodyweight alone is an adequate initial stimulus to achieve a positive response. However, plateaus will undoubtedly occur and therefore an increase in intensity would lead to preferential adaptations, which is supported by Pollard et al.(2019), where the greater training intensity resulted in the greatest adaptations to eccentric hamstring strength and BF_{LH} FL. Consistent with interventions observing BF_{LH} FL, concentric resistance training intensity was between 60 – 83% of 1RM load (Bourne, Duhig, et al., 2017; Duhig, et al., 2019), for both a knee dominant and hip dominant tasks (Bourne, Duhig, et al., 2017; Duhig, et al., 2019). The percentage working intensities highlighted are similar to those which would be prescribed for optimal hypertrophy leading into strength-based intensities for assistance exercises, as prescribed by the National Strength and Conditioning Association (NSCA) (Sheppard and Triplett, 2016). Although, it could also be presumed that even greater loads (>85% 1RM) may lead to further increases in eccentric hamstring strength as they are more consistent with strength-based intensities (Sheppard and Triplett, 2016). Although with hip dominant exercises such as the Romanian deadlift (RDL) there may be a risk-reward trade off, as loading >85% could be very demanding, particularly at high-loads (>85% 1RM), some pre-conditioning or history of heavy strength training would be crucial. Duhig et al. (2019) utilised a unilateral concentric leg curl exercise, which also produced moderate increases in eccentric hamstring strength – despite differences in architectural adaptations, potentially signifying a different adaptative response, including; other structural or neural adaptations and the potential strengthening of connective tissue of the distal portion of the hamstring MTU (Franchi, Reeves, & Narici, 2017; Heinemeier et al., 2007; Higbie, et al., 1996; Jakobsen et al., 2017).

The frequency of training ranged from 1-3 x/week, with a number of NHE interventions employing 1 per week frequency (Askling, et al., 2003a; Pollard, et al., 2019; Presland, et al., 2018). Despite these low frequencies, moderate to very large increases in eccentric hamstring strength were still observed (Askling, et al., 2003a; Pollard, et al., 2019; Presland, et al., 2018). This could highlight how a low frequency approach to using the NHE could be utilised within practice, this could open the prospect of using a range of methods across the training week (i.e., day 1: NHE, day 2: hip dominant concentric-eccentric and day 3: sprinting). There was a large range of training volumes utilised across all resistance training interventions ranging from 21.3 – 120 repetitions per week, whist volume per sessions ranges from 8 – 60 repetitions. Although, as Presland et al. (2018) prescribed a low volume approach had the greatest positive effect on eccentric hamstring strength in comparison to high volume equivalent, this could be an effect of an initial higher, matched volume control period, where a large rebound of supercompensation could have occurred (Haff, 2016; Hyldahl, Chen, & Nosaka, 2017). Despite the range of prescribed volumes being effective at increasing eccentric hamstring strength, low frequency/volume approach could be utilised, where multiple modes are implemented within the training week.

2.7.3.4 Consideration of the eccentric hamstring strength assessments across interventions

Within the current systematic review, there were two primary methods of assessing eccentric hamstring strength, the gold standard, lab-based fixed velocity assessment (i.e., isokinetic), while the alternative is a field-based measure (i.e., Nordbord). To date, a single study has shown that there is minimal agreement between a Nordbord-type device and isokinetics (Wiesinger, et al., 2019), with both devices reflecting eccentric hamstring strength in divergent ways (Wiesinger, et al., 2019). This is unsurprising as the methods of assessment are essentially different, with diverse lever and moment arms in addition to segment weights, utilised with each device (Figure 2-13).

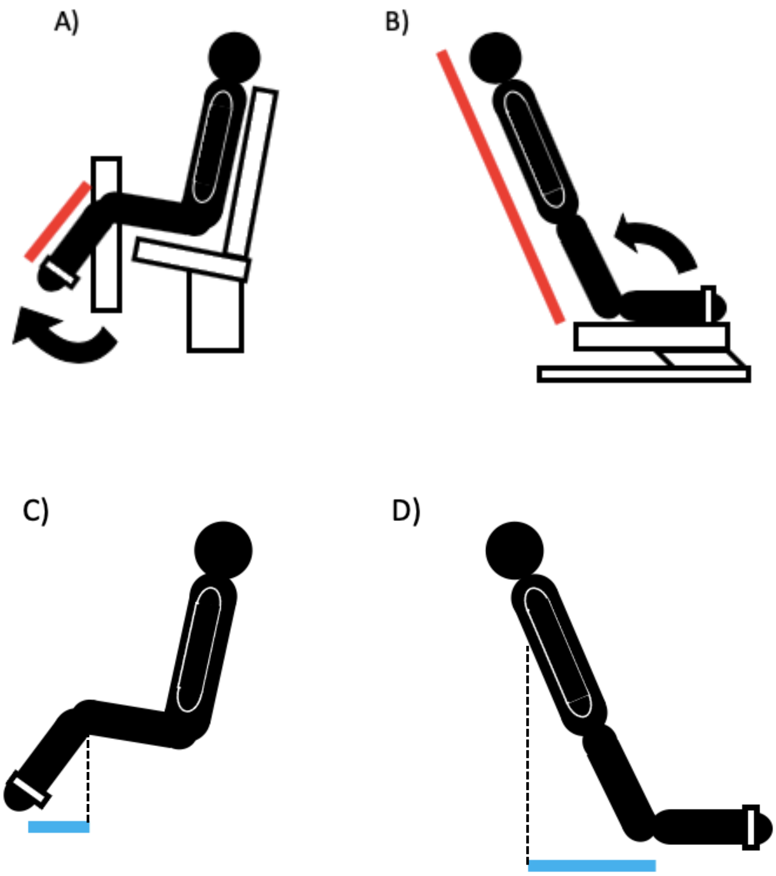


Figure 2-13 Representations of difference between eccentric hamstring assessment with differences in segment length (red) and moment arms (blue) A and C) Isokinetic, B and D) Nordbord. The moment arm during Isokinetic assessments is fixed to shank length, whereas the moment arm will increase in the NHE with further knee extension.

Across the studies that have used isokinetic assessments, a series of isokinetic velocities have been utilised (15-240 deg/s), despite the range of velocities all have resultant increases in eccentric hamstring strength, with minimal differences between slow-fast isokinetic velocities (Alt, et al., 2018; Iga, et al., 2012). Across the studies both absolute and relative measures of eccentric hamstring have been reported – although this made minimal difference on the observed magnitudes (Delahunt, et al., 2016; Suarez-Arrones, et al., 2019). It could however impact the usefulness of studies which have reported relative measures alone, as changes in relative strength could have been a result of changes in body mass rather than strength, which is commonly not reported with the studies. In addition, the Nordbord is more dependent upon body mass than isokinetic strength (up to 24%) (Buchheit, et al., 2017), although further research is warranted on the relationships between body mass and measures of eccentric hamstring strength (Nordbord and isokinetic) and athletic populations (Roe, Malone, et al., 2018).

2.7.4 Conclusions

For an optimal modality, frequency and training volume it is difficult to come to a conclusion with the present literature; as various modalities, volumes, frequencies and intensities all lead to positive improvements in eccentric hamstring strength. However, as HSIs are a frequent problem within team-based sports, with congested fixture and training schedules, a dose which achieves the positive adaptations to eccentric hamstring strength – while minimising muscle soreness (DOMS) and fatigue may be considered optimal. This could potentially include a holistic approach, where supra-maximal eccentric exercises, traditional concentric-eccentric resistance training and sprinting make part of the weekly training process. However, within the current research only single exercise interventions have been utilised – which is unlikely to happen within practice, where an encompassed approach to training is more realistic. Furthermore, there remains a number of methodological questions, including sprinting volumes (optimal distance per repetition, role of the acceleration phase).

2.8 Electromyography

Electromyography (EMG) is frequently used to assess relative muscle activation in a range of athletic and occupational tasks (Ball and Scurr, 2011; Chapman, Vicenzino, Blanch, Knox, & Hodges, 2006; Fauth, Petushek, Feldmann, Hsu, & Garceau, 2010; C. Hansen, Einarson, Thomson, & Whiteley, 2017; Higashihara, Nagano, Ono, & Fukubayashi, 2015; Higashihara, et al., 2016; Higashihara, Nagano, Ono, & Fukubayashi, 2018; Higashihara, Ono, Kubota, Okuwaki, & Fukubayashi, 2010; Onishi et al., 2002). Farina et al. (2004) described that EMG amplitude is the net motor unit activity and reflects the recruitment and discharge rates of active motor units. This allows for practitioners to understand the neuromuscular requirements of muscle actions including muscular contributions and control patterns that will allow for appropriate technique modification and inform training processes (Ball and Scurr, 2013).

With a focus on hamstrings, there is a wealth of EMG task-based literature, identifying a variety of metrics; including peak and mean EMG signal amplitudes (Bourne, Williams, et al., 2017; Comfort et al., 2017; Contreras, et al., 2015; Contreras, Vigotsky, Schoenfeld, Beardsley, & Cronin, 2016a; Higashihara, et al., 2015, 2016, 2018; Tsaklis et al., 2015). Further research has included the use of time-related metrics including the rate of rise (RoR) of EMG amplitude at various time intervals (e.g., 30, 50, 100 and 200 ms) and time-integrated EMG (iEMG) (Aagaard, Simonsen, Andersen, Magnusson, & Dyhre-Poulsen, 2002; Barry, Warman, & Carson, 2005; Jenkins et al., 2014; Kyrolainen, Avela, & Komi, 2005; Kyrolainen et al., 2005). Although each of the measures identified expresses slightly different neuromuscular contributions, they are in fact complimentary and together provide a more comprehensive view on the neuromuscular characteristics of various exercises or tasks. The characteristics that can be distinguished include: the peak requirement to the task (peak EMG), the activation velocity of the required task (RoR) and the overall contribution of the muscle to the task (iEMG).

Neural characteristics contribute to the mechanical output (e.g., absolute force production) and efficiency rates (ratio of work performed to energy expenditure) of muscles (Duchateau and Baudry, 2014). Amplitudes of muscle activation derived from surface EMG have demonstrated that; isometric, concentric and eccentric muscle actions, produce meaningfully different intensities (Aagaard et al., 2000; Amiridis et al., 1996; Duchateau and Baudry, 2014; Farina, et al., 2004; Tesch, Dudley, Duvoisin, Hather, & Harris, 1990; Westing, Cresswell, & Thorstensson, 1991). Generally, eccentric muscle actions produce a significantly lower EMG amplitude than both concentric and isometric muscle actions at the same relative intensities

(Farina, et al., 2004; Fauth, et al., 2010; Grabiner and Owings, 2002; Kim, Thompson, & Hornby, 2015; Madeleine, Bajaj, Søgaard, & Arendt-Nielsen, 2001; M. P. McHugh, et al., 2002; Ono, Okuwaki, & Fukubayashi, 2010; Timmins et al., 2014). This concept has been challenged within the literature with some researchers identifying that between modes of muscle action, there is no significant difference in EMG amplitudes at the same intensity (Babault, Pousson, Michaut, Ballay, & Hoecke, 2002; Baudry, Klass, Pasquet, & Duchateau, 2007; Carroll et al., 2019; Duclay and Martin, 2005; Duclay, Pasquet, Martin, & Duchateau, 2011; Hahn, Hoffman, Carroll, & Cresswell, 2012; Linnamo, et al., 2003).

By drawing on the concept of increasing EMG amplitude, when moving between eccentric, isometric and concentric muscle actions (Duchateau and Baudry, 2014), some researchers have attempted to utilize EMG amplitudes alone to determine the specific muscle actions of varying muscle groups (Ekstrom, Donatelli, & Carp, 2007; Jonhagen, Halvorsen, & Benoit, 2009; Ono, Higashihara, & Fukubayashi, 2011). Although determining the muscle action condition from EMG amplitude alone would be of interest, it cannot however, distinguish with any validity or accuracy specific muscle actions that occur through a movement or task (e.g., maximal sprinting) (Bourne, Williams, et al., 2017; Higashihara, et al., 2015, 2018; Schache, et al., 2013; Tsaklis, et al., 2015; Van Hooren and Bosch, 2017a, 2017b). Amplitude of EMG can tell a story with regards to muscle excitation, the absolute intensity of action (e.g. 50% 1RM or 120% 1RM) determines the amplitude values regardless of the muscle action performed (i.e. 50% concentric amplitude ≈ 120% eccentric amplitude) (Ono, et al., 2010). The use of *in-vivo* technologies (i.e., US) is the only method currently available, that can accurately determine the specific muscle action which is occurring alongside EMG amplitude (e.g., using US to determine muscle action based on changes in muscle fascicle and tendon length).

2.8.1 Assessment

There are two main methods used to assess muscle activation via EMG including intramuscular EMG and surface EMG. Intramuscular EMG involves the insertion of fine wire electrodes into the muscle (s) under investigation (Onishi, et al., 2002), whereas surface EMG involves the application of electrodes above the muscle under investigation, with previous research identifying optimal electrode placement guidelines (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000; Rainoldi, Melchiorri, & Caruso, 2004). Given that intramuscular EMG is an extremely invasive procedure, posing difficulties when assessing EMG amplitude during dynamic tasks, as well as the high levels of reliability that can be achieved when using surface EMG, it will not be examined for this thesis (Burden, Trew, & Baltzopoulos, 2003; Fauth, et al., 2010; Larsson, Karlsson, Eriksson, & Gerdle, 2003).

Bipolar EMG electrode configurations are commonly used by researchers, whereby two electrodes with a small inter-electrode distance and a reference electrode are placed upon the muscle belly and a passive structure respectively. This electrode configuration is advantageous as the common noise received from the two electrodes, can be eliminated allowing for a cleaner EMG signal for analysis. Electrodes normally contain a silver or silver-chloride backing (Ag-AgCl), this is used to decrease the impedance of the electric currents (Duchêne and Gouble, 1993). Electrode cables are then attached to the electrodes, with the data usually being collected in an analogue form, requiring pre-amplifying as well as low and high pass filtering to remove unwanted artefacts (e.g., noise artefacts, including, movement of the cables) (Gerdle, Karlsson, Day, & Djupsjöbacka, 1999). Method alterations could be implemented including taping and securing of loose electrode cables, in order to reduce the unwanted artefacts within EMG data, specifically effective for tasks involving a high degree of movement. With advancements in technology wireless-based systems including the attachment of small amplifiers to the skin have come into use. These provide practitioners an ability to reduce random artefacts from wire movements which may produce type 1 errors during analysis.

The minimum sampling of most EMG data is 250 Hz, but most units have the capability to sample upwards of 2000 Hz. Ives and Wigglesworth (2003) found that with lower frequencies (250 - 500 Hz), there was a significant difference in peak EMG amplitude compared to the greatest achievable sampling frequencies (6000 Hz), with no significant difference between 1000-, 3000- and 6000 Hz for peak, average and total EMG amplitudes across a variety of movement types, including isometric, submaximal concentric, maximal velocity concentric and concentric under fatigue. Suggestions were made prescribing that 1000 Hz is a sufficient and a functional minimum sampling frequency for the collection of surface EMG data (Ives and Wigglesworth, 2003), and that sampling at less than this reduces the usefulness of the raw EMG data (Ives and Wigglesworth, 2003). Under sampling becomes particularly important when looking at onset of activation, which could be used for identification of different temporal phases of a movement.

2.8.2 Normalisation

Normalisation involves the comparison of the task EMG signal to a reference EMG value obtained during reproducible conditions (Albertus-Kajee, Tucker, Derman, & Lambert, 2010; Ball and Scurr, 2011, 2013; Burden, 2010). Rescaling of the raw task EMG data can be achieved via the equation.

$$Normalised\ EMG = (Task\ EMG \div Reference\ EMG) \times 100$$

Equation 2-2 Normalization of task EMG.

Rescaling EMG allows for improvement of the intra- or inter-individual reliability, providing a representative measure of muscle activation during a task as well as permitting for comparison of activity between different muscles, across time and between individuals (Albertus-Kajee, et al., 2010; Ball and Scurr, 2011, 2013; Burden, 2010). A review by Burden (2010) identified eight different methods of normalization within the literature (Table 2-10).

Table 2-10 Methods of EMG Normalization, adapted from Burden et al., (2010)

Method Name	Acronym	Methodology
Mean Task	$Mean_{task}$	Mean EMG from the task under investigation
Peak Task	$Peak_{task}$	Peak EMG from the task under investigation
Submaximal isometric action	SubmaxISO	Peak EMG from a submaximal isometric voluntary action
Submaximal dynamic action	SubmaxDyn	Peak EMG from a submaximal dynamic voluntary action
Arbitrary angle MVIC	ArbMVIC	Peak EMG from an MVIC action obtained from an arbitrary mid-range joint angle
Angle specific MVIC	SpecMVIC	Peak EMG from an MVIC action obtained from the same joint angle or muscle length as the task
Angle specific maximal dynamic voluntary action	SpecDyn	Peak EMG from a dynamic action obtained from the same muscle action, joint angle or muscle length as the task
Specific maximal isokinetic action	SpecIK	Peak EMG from an isokinetic action obtained from the same muscle action, joint angle, muscle length, and angular velocity or rate of change in muscle length as the task

One issue which is not clearly defined within the review (Burden, 2010), is what is the difference between Spec Dyn and $Peak_{task}$ normalisation methods? For instance, if the task under investigation is a series of running trials or resistance exercise, how would it be possible to replicate the actions, joint angles or muscle length of the task for the SpecDyn method, without utilizing the peak EMG from the task that is under investigation itself? High levels of reliability have been identified for all methods of normalization, with an ArbMVIC method being shown to be more reliable than all other methods of normalization (Burden, 2010; Burden, et al., 2003; Lehman, 2002; Rouffet and Hautier, 2008).

One limitation when using any MVIC method, however, is that it is difficult to determine if the participant is producing a true maximal contraction without twitch interpolation (Burden, 2010), although this type of analysis is not without its own short comings. What is unclear however, is if an MVIC can be used to normalize dynamic task EMG e.g., running. Ball and Scurr's (2013) recommendations on normalizing EMG data for dynamic tasks describes that the normalization method utilized should involve an action similar to the task under investigation (e.g., SubmaxDyn, $Mean_{task}$, $Peak_{task}$ or SpecDyn), which include a maximal effort (e.g., 20 m maximal sprints). The rationale for this is that it would incorporate the same neural conditions as the measured activity (i.e., motor unit recruitment, rate coding and synchronization) (Klein, Peterson, Ferrell, & Thomas, 2010).

A further consideration that should be made when using normalization techniques, is that the intensity of EMG amplitude is dependent upon a number of mechanical factors including: the force produced, joint angle, muscle length, degree of synergistic action, relative location of fast and slow twitch muscle fibres, action velocity and activation/deactivation kinetics. Although, Burden (2010) discusses the effect of joint angle on MVICs, describing that there is little effect on the MVIC with differences of joint angle and muscle length. However, joint angular position could have a key influence upon EMG intensity, because of changes in muscle length altering the efficiency of muscle force generation, due to the length-tension relationship (Kaufman, An, & Chao, 1989; Rassier, MacIntosh, & Herzog, 1999). As it is generally accepted that EMG amplitude increases with the force of contraction, under isometric and isotonic conditions (Basmajian and De Luca, 1985; Karlsson and Gerdle, 2001; Lawrence and De Luca, 1983; Moritani and Munro, 1987), however, it is disputed if there is a linear or non-linear increase in EMG amplitude (Kuriki et al., 2012).

Normalization processes therefore may only be suitable for specific research designs, as in when comparing between groups of individuals, or using EMG on multiple occasions (Burden, 2010). Burden (2010) explained that a $\text{Mean}_{\text{task}}$ method would be preferential when normalizing EMG data to maximize the reduction in variability between participants (Bolgla and Uhl, 2007; Burden, 2010), with $\text{Peak}_{\text{task}}$ also being preferential over other methods of normalization (Bolgla and Uhl, 2007; Burden, 2010). However, if using a within-group study design and EMG is assessed upon a single occasion, where the intensity of an exercise/exercises is under investigation, a normalization process may not provide any further information of the EMG characteristics (Burden, 2010), as it would not allow for comparisons between magnitude or patterning between muscles and task variations (Burden, 2010). Additionally, it could mask task intensity, as varying methods can produce meaningfully different EMG values (Ball and Scurr, 2013). Therefore, for specific research designs, normalization could be an irrelevant process – where a more accurate reflection of task intensity and patterning could be provided from non-normalized EMG.

2.8.3 Filtering

After data collection is complete and the raw data has been exported, it requires signal averaging. This is commonly performed via the equation of root mean square (RMS).

$$RMS = \sqrt{(V^2/P)}$$

Equation 2-3 Root Mean Square equation to Filter EMG

Where V = voltage data across a given window and P = total number of data points. This method of data processing is required as raw EMG amplitude varies above and below an absolute zero in a random nature (Figure 2-14). Therefore, rectifying the raw EMG data signal via the RMS calculation allows for further analysis with only positive values considered (Figure 2-15).

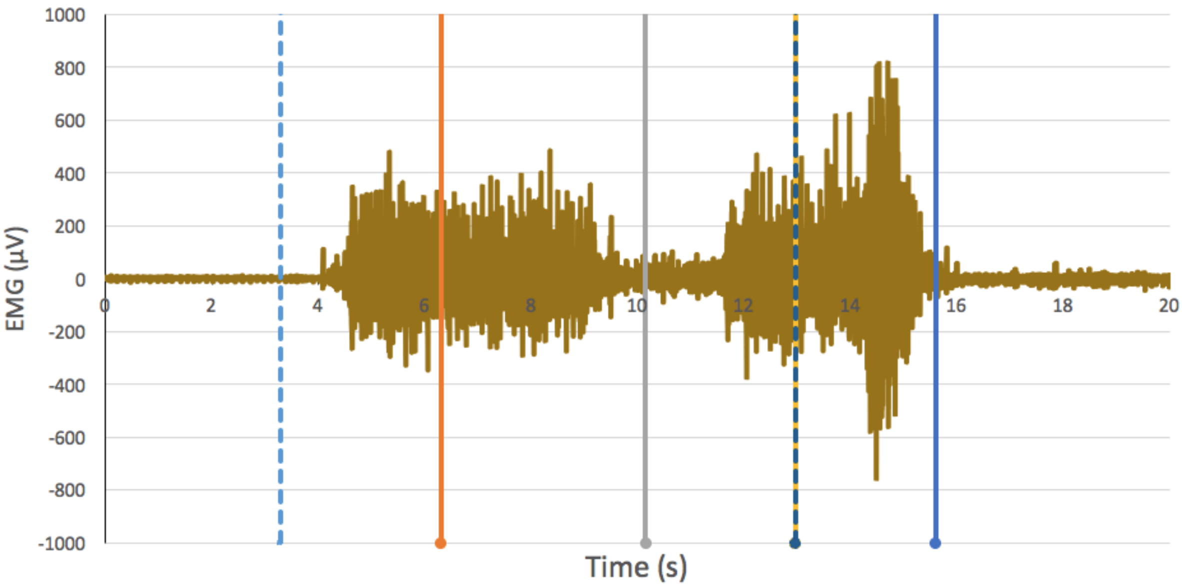


Figure 2-14 Typical raw EMG trace of the bicep femoris during the glute-hamstring raise exercise.

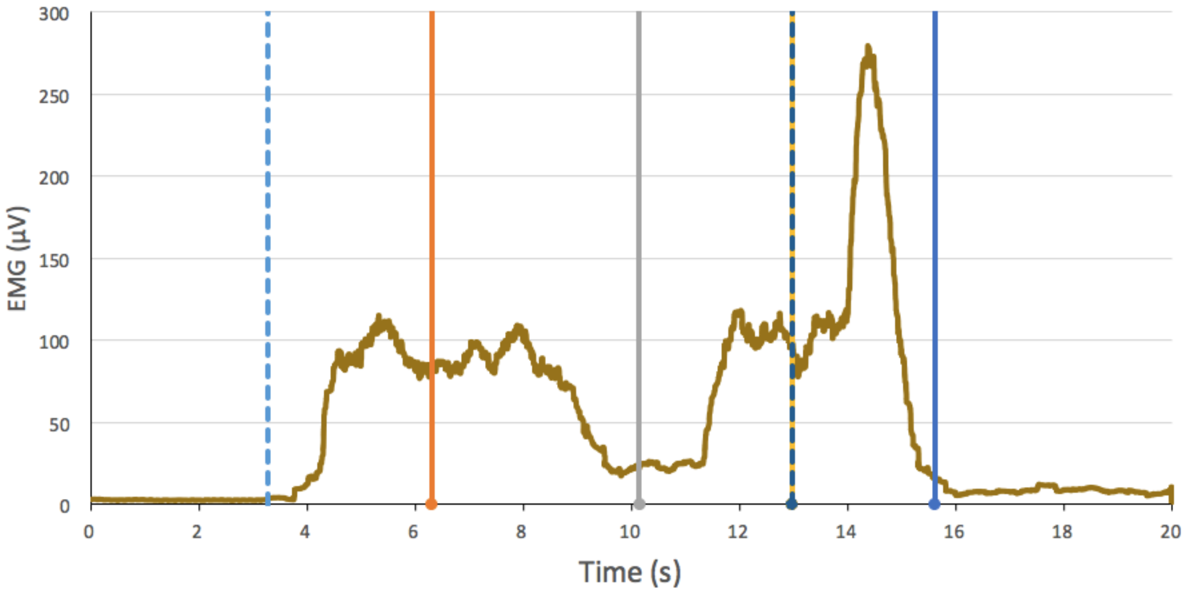


Figure 2-15 Typical rectified EMG trace of the Bicep Femoris during the glute-hamstring raise exercise.

The average window length must be consistent, however, as the variability when using different windows lengths can be fairly large (Bamman, Ingram, Caruso, & Greenisen, 1997). This is not unexpected as larger windows (e.g., ≥ 500 samples) would offer a greater smoothing effect upon the data, compared to smaller average window lengths (e.g., ≤ 100 samples). Further to this, not only should the average window length be defined, but it should also be specific to the research objective (i.e., if performing more dynamic tasks including higher velocity movements (e.g., sprinting) use a shorter window length).

When testing the MVIC of a given muscle, because this is a static event and observing a peak voltage reading, larger average window lengths could be used as long as this window length remains the same for further testing. Greater levels of reliability and decreased variability of MVIC results have been shown to occur with larger sampling windows (500 and 1000 samples) when compared to windows of smaller lengths (100 and 200 sample windows) (Bamman, et al., 1997). Contrastingly greater EMG values are seen with the smaller window lengths, with 100 samples being found significantly greater than 200, 300, 400 and 500 samples, for MVIC testing (McLean, Chislett, Keith, Murphy, & Walton, 2003). This is understandable as larger windows offer a greater filtering effect on raw EMG data. This is represented within the literature, with some researchers having previously used windows of ≥500 samples in length (Bamman, et al., 1997; Bolgla and Uhl, 2007; Clancy and Hogan, 1997; Kollmitzer, Ebenbichler, & Kopf, 1999).

Smaller windows of 20-200 samples have been used within the research when attempting to compare EMG results, specifically when exploring dynamic movements (e.g. strength exercises, low velocity movements and high velocity movements) (Albertus-Kajee, et al., 2010; Babault, et al., 2002; Ball and Scurr, 2011; Bourne, et al., 2016; Brandon, Howatson, & Hunter, 2011; Burden, et al., 2003; Ditroilo, De Vito, & Delahunt, 2013; Ekstrom, et al., 2007; Fauth, et al., 2010; Giorgio, Samozino, & Morin, 2009; Jonhagen, et al., 2009; Kollmitzer, et al., 1999; Maenhout, Benzoor, Werin, & Cools, 2016; M. P. McHugh, et al., 2002; Morin, et al., 2015; Opar, Williams, et al., 2013a; Rouffet and Hautier, 2008; Zebis et al., 2013). The rationale behind the use of the smaller window length is that these studies compare dynamic movements, which if performed with longer window lengths (≥ 500 sample window) may have disguised phases of the cyclical action of these movements. This is of particular importance when trying to observe changes in very discrete movements (e.g. differences in EMG at different knee angles), Jakobsen and colleagues (2013) studied muscle activity during strength training exercises, specifically identifying changes in EMG activity at different knee angles (0° – 10°, 10° - 20°, . . . 80° - 90°). The movement time between these angles was identified as being 100 – 200 ms constant but the researchers used a filter length of 500 ms, which would filter data over multiple angles and thus mask key differences between each angle.

Unfortunately there are a series of studies that have mentioned the use of an RMS equation to filter the data but have not followed on to state the windows they had used to calculate their resultant data (Comfort, et al., 2017; Fernandez-Pena, Lucertini, & Ditroilo, 2009; Higashihara, et al., 2016, 2018; Iga, et al., 2012; Liebenberg et al., 2011; Madeleine, et al., 2001; Munera et al., 2017; Opar, et al., 2015; Schoenfeld et al., 2015; Timmins, et al., 2014; Tsaklis, et al., 2015). Another limitation when reporting EMG data, is reporting the sampling window as a unit of time (e.g., 500 ms) and then not proceeding to identify the sampling frequency of the EMG measurement device (Guex, et al., 2012), as the sampling frequency will affect the actual sampling window length (i.e. exact number of data points used within the RMS filtering equation). Where this poses a problem is when looking to compare between studies and when trying to replicate studies, because of the large differences in EMG amplitude seen between different RMS filter window lengths. Therefore, all future studies should identify both the sampling frequency of the EMG measuring device and the RMS filter window length in either number of samples or the specific time window. Additionally, when using RMS calculations to assess MVIC amplitude, for normalization purposes, it would be suggested to use the window length specific to the minimum time of the measured action (i.e., the time constant between changes in knee flexion angle or time taken to perform the swing phase of running). As differences between these may affect the normalization percentages by inflating the normalized amplitude percentages.

2.9 Ultrasound

2.9.1 What is ultrasound?

Ultrasonography has been used within clinical practice since the early 1950s (Kane, Grassi, Sturrock, & Balint, 2004; Loram, Maganaris, & Lakie, 2006; Whittaker et al., 2007), however, with constant improvements made in the imaging quality, later uses for US include the measurement of muscle architecture characteristics (Abe, Fukashiro, Harada, & Kawamoto, 2001; Abe, et al., 2000; Ando et al., 2018; Behan et al., 2018; Bodine et al., 1992; Kawakami, Abe, & Fukunaga, 1993; Kumagai, et al., 2000; Timmins, Bourne, et al., 2016; Timmins, et al., 2015). Within the literature US measurements have included muscle CSA, MT, PA and muscle FL, across a variety of muscles (Abe, et al., 2001; Abe, et al., 2000; Ando, et al., 2018; Behan, et al., 2018; Bodine, et al., 1992; Kawakami, et al., 1993; Kumagai, et al., 2000; Timmins, Bourne, et al., 2016; Timmins, et al., 2015).

The measurements derived from sonographic still images at rest, have been used by researchers to assess associations with performance (Abe, et al., 2001; Abe, et al., 2000; Kawakami, et al., 1993; Kumagai, et al., 2000; Suchomel and Stone, 2017) and injury risk (Opar, et al., 2012; Timmins, Bourne, et al., 2016; Timmins, et al., 2015; Timmins, Shield, Williams, Lorenzen, et al., 2016), as well as identifying any changes or adaptations in muscle architecture from aging (Narici, Maganaris, Reeves, & Capodaglio, 2003), the occurrence of injuries (Nagano, et al., 2015; Timmins, et al., 2017) and training (Blazevich, Gill, Bronks, & Newton, 2003; Bourne, Duhig, et al., 2017; Bourne, et al., 2018; Duhig, et al., 2019; Kawakami, et al., 1993; Presland et al., 2017; Presland, et al., 2018; Timmins, Ruddy, et al., 2016). More recently sport scientists have utilized the live imaging and recording options available, to observe dynamic changes of FL *in-vivo*, with varying degrees of success (Bohm, Marzilger, Mersmann, Santuz, & Arampatzis, 2018; Cataneo, 2018; Kellis, 2018; Raiteri, 2018; Stubbs et al., 2018).

2.9.2 Assessment of ultrasound at rest

Muscle architecture assessment relates to the determination of muscle FL, this is of particular interest within the BF_{LH} due to the strong relationship between decreased risk of HSI with an increased BF_{LH} FL (Timmins, Bourne, et al., 2016; Timmins, Ruddy, et al., 2016; Timmins, et al., 2015). Assessment of muscle architecture can be performed using MRI, although there are a number of limitations to this method reducing its viability, including the high expense and long duration of assessment (Damon, Ding, Anderson, Freyer, & Gore, 2002). Therefore, US is more commonly used as in contrast to MRI the assessments are fast, the US devices are relatively inexpensive when compared to MRI machines and the high reliability that can be achieved with US assessment.

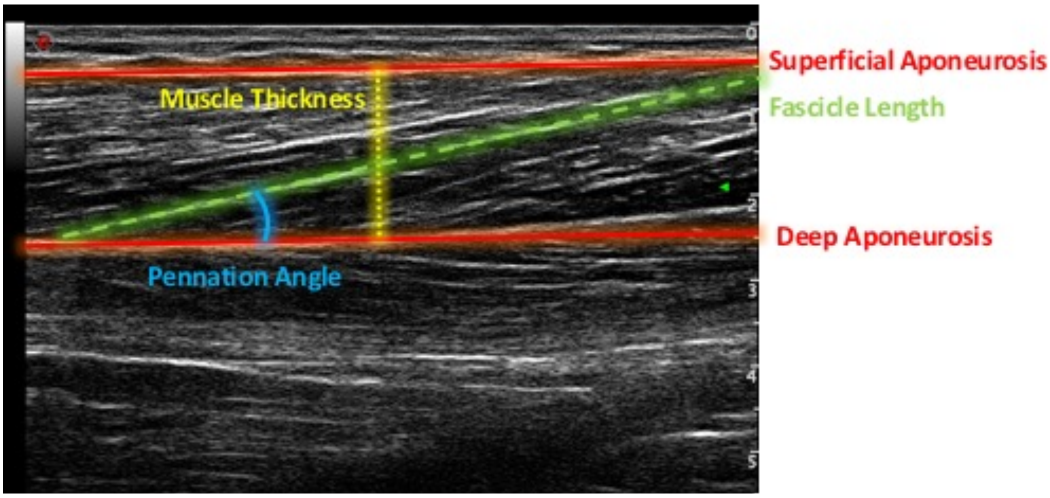


Figure 2-16 Resting image of BF with annotations depicting the aponeurosis, muscle thickness, pennation angle and the estimated fascicle length for Equation 2-4.

For the hamstrings, assessment of BF_{LH} muscle architecture requires the collection of two-dimensional (2D) images of the muscle (Figure 2-16). Where both superficial and deep aponeuroses can be identified with a parallel orientation, the perpendicular distance between these two points is defined as the MT (Timmins, Bourne, et al., 2016; Timmins, Ruddy, et al., 2016; Timmins, et al., 2015). A fascicle of interest which can be clearly identified connecting to the deep aponeurosis can then be marked, and the angle between the two landmarks measured and given as the PA (Timmins, Bourne, et al., 2016; Timmins, Ruddy, et al., 2016; Timmins, et al., 2015). As the name suggests, FL is the given length of a fascicle between the two aponeuroses. However, there is a caveat particularly when assessing BF_{LH} FL, in that the entire fascicle is not always visible particularly when assessed with probes that possess a smaller FOV, with previous research typically using probes of <6.5 cm in length (De Oliveira, et al., 2016; Franchi, Fitze, et al., 2019; Freitas, et al., 2018; Kellis, et al., 2009; Lima, Carneiro, Alves, Peixinho, & De Oliveira, 2015; Pimenta, et al., 2018). Therefore, probes that have a greater FOV will potentially be more accurate, when attempting to define BF_{LH} FL.

As the entire FL is typically not visible in a single US image, to assess the total BF_{LH} FL it therefore requires estimation (Pimenta, et al., 2018; Timmins, et al., 2015). Within the research, estimation of total BF_{LH} FL from a single image has been performed, typically utilizing one of three equations using visible architectural properties (Equation 2-4, 2-5 & 2-6), with similar applications of trigonometry. Of the three equations, the simplest (Equation 2-5) estimates the entire FL from measurements of MT and PA, as demonstrated in Figure 2-16. However, this equation ignores the possibility of fascicle or aponeurosis curvature (aponeurosis angle (AA), therefore a second equation (Equation 2-4) that overcomes these draw backs may be better suited at estimating total BF_{LH} FL (Figure 2-17). More recently, a third equation (Equation 2-6) used to estimate BF_{LH} FL from a single US image, this equation measures the visible BF_{LH} FL to a distal point, then using a comparable process to equation 2-5 estimating the remaining un-measurable portion of the BF_{LH} fascicle (Figure 2-17). Similar to Equation 2-5, this process ignores the potential for fascicle or aponeurosis curvature however removes the large degree of estimation, as it only marginally estimates the un-measurable portion, which depending on the probe length could be minimal. However, no research to date has looked to comparing the values between the three-estimation equations used within the research.

$$FL = sin(AA + 90deg)\ x\ MT/sin(180deg - (AA + 180deg - PA)$$

Equation 2-4 Criterion method of fascicle length estimation.

$$FL = MT/(sin\ (PA))$$

Equation 2-5 Fascicle length estimation using basic trigonometry.

$$FL = L + (h\ \div\ sin\ (\beta))$$

Equation 2-6 Fascicle length estimation partial measure equation

Where L is the observable fascicle length, h is the perpendicular distance between the superficial aponeurosis and the fascicles visible end point and β is the angle between the fascicle and the superficial aponeurosis.

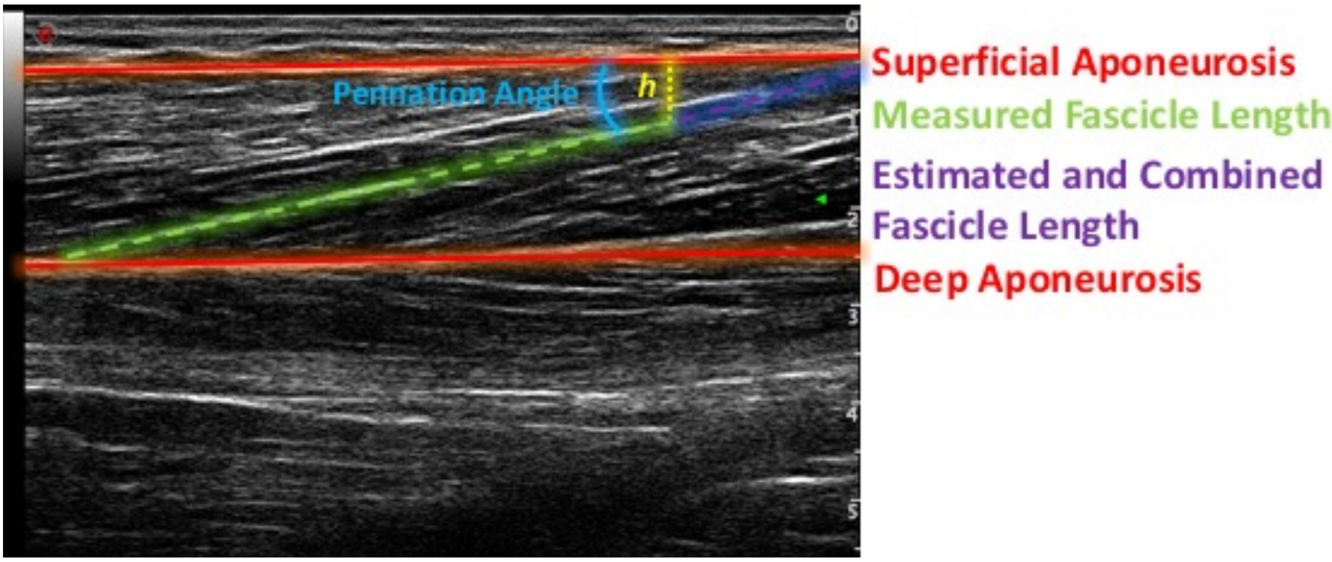


Figure 2-17 Resting image of BF with annotations depicting the third method of estimating BF FL, where h is the perpendicular distance between the superficial aponeurosis and the fascicles visible end point.

Previous studies have identified excellent reliability of BF_{LH} muscle architecture assessment, with ICCs of > 0.90, CV percentages of between 2.1 – 3.4 % and standard error measurement (SEM) or typical error measurement (TEM) percentage values of < 10.5% reported across the literature for MT, PA and FL for the BF_{LH} (De Oliveira, et al., 2016; Franchi, Fitze, et al., 2019; Freitas, et al., 2018; Kellis, et al., 2009; Lima, et al., 2015; Pimenta, et al., 2018). Although the test-retest reliability described by De Oliveira and colleagues (2016) was markedly lower (0.81 – 0.92) than what was reported by Timmins et al. (2015). Two possible explanations for this, the first being the difference in the method utilized in to find mid-point of the BF_{LH}, with Timmins et al. (2015) using the ischial tuberosity and De Oliveira et al. (2016) using the greater trochanter. Another possible explanation being that the sample used De Oliveira et al. (2016) included female participants who possess a greater fat composition, which may negatively influence image quality.

As US images are generally analysed manually, it has been proposed that this can introduce types of error and experimenter bias (Cronin, Carty, Barrett, & Lichtwark, 2011). This could potentially reduce the reliability and repeatability of measurements due to the nature of the subjective assessment (Aeles et al., 2017). Therefore, it is crucial to not only determine the ability of each assessor to reliably collect and analyse muscle architecture, in addition to, the TEM, which is the variation from measurement to measurement (Hopkins, Schabort, & Hawley, 2001). This becomes especially important, when attempting to determine what could be classified as meaningful changes or adaptations in muscle architecture, without which any changes seen could be perceived as being the internal variation within the test (i.e., measurement error).

An alternative method of estimating or measuring BF_{LH} FL from a single image includes linear extrapolation of a visible fibre. This proposed method is comparable to Equation 2-5, whereby the visible fascicle is measured, and the remaining is estimated by the line of the visible fascicle being extended. Additionally, more in-depth US methodologies have been proposed, that have the potential to image an entire fascicle, these include dual-probe and extended FOV methodologies. Recent research has looked to compare between fascicle measurements from a single image and varying extended FOV methods (linear and non-linear) (Pimenta, et al., 2018). Pimenta and colleagues (2018) identified that estimations of BF_{LH} FL made from a single image, were significantly greater with moderate ES than both linear and non-linear extended FOV methods. Additionally, the single image technique significantly underestimated pennation angle when compared to linear extended FOV methods, with a non-significant underestimation when compared to a non-linear FOV method (Pimenta, et al., 2018). The authors proposed that these differences could be in part due to a single image not accounting for a full identification of the superficial aponeurosis trajectory, or fascicle curvature (Pimenta, et al., 2018).

Using a single image technique (equation 2-5), 35.4 ± 7.0% of the BF FL was estimated (Pimenta, et al., 2018). However, given that FLs of >12 cm has been reported within the literature (Timmins, Bourne, et al., 2016), it would be impossible for practitioners to image anywhere near 50% of FL specifically with the 6-cm probe reported by most researchers, requiring practitioners to estimate a larger proportion. Indicating that for highly trained athletes, i.e., those with an expected greater BF_{LH} FL, a greater error rate could be expected. Therefore, it would be prudent for the collection of accurate images, that extended FOV methods be adopted within research and practice. However, these techniques are more time-consuming for both data collection and analyses; therefore, would be impractical within the field. A potential alternative to extended FOV methods that would be more practical, could be to utilise probes which possess a greater field of view ~ 9 cm and using single image technique. However, to date limited research has been performed investigating the ability of longer probes (~ 9 cm) to assess BF_{LH} FL (Behan et al., 2019), therefore requiring further investigation.

2.9.3 Dynamic ultrasound assessment

To achieve maximal performance within athletic tasks, such as running, jumping, cycling and resistance exercises, the in-vivo dynamics of skeletal muscle are obviously integral for optimum performance (Beaumatin et al., 2018; Cataneo, 2018; Cronin, et al., 2011; Drazan, Hullfish, & Baxter, 2019; Earp, Newton, Cormie, & Blazevich, 2017; Farris and Lichtwark, 2016; Heroux, Stubbs, & Herbert, 2016; Kellis, 2018; Lai, Biewener, & Wakeling, 2019; Lai et al., 2015; Lichtwark and Wilson, 2006; Loram, et al., 2006; Peter, Hegyi, Finni, & Cronin, 2017;

Raiteri, 2018; Rubenson, Pires, Loi, Pinniger, & Shannon, 2012; Zhou, Li, Zhou, & Zheng, 2012). Although accurate imaging can be of an increased difficulty, particularly at higher movement velocities, the information that could be attained can improve practitioners understanding of exercise and training prescription.

Under active conditions, eccentric actions incorporate a resistance against a force – i.e., resisted lengthening, where the aim is to resist rapid deformation. Within resisted knee flexion, where the quadriceps are required to produce an eccentric action, fascicles of the quadriceps are required to lengthen (Ando, et al., 2018; Guilhem, Cornu, Guével, & Guevel, 2011; Ishikawa, Finni, & Komi, 2003). During an active eccentric muscle action for the quadriceps, a similar magnitude of lengthening is achieved regardless of the isokinetic velocity utilised (Finni, Ikegawa, Lepola, & Komi, 2003). However, the strategy in the way they reach the magnitude is different; isokinetic velocity appears to dictate the magnitude fascicle lengthening within the early phases of movement (resisted knee flexion (Finni, et al., 2003)), potentially as a result of the large difference in PA between the application of varying velocities. With slower velocities (60 deg/s), there in an increased PA across the majority of the ROM, until the final degrees of motion where PA was found to be similar, which may explain the consistent magnitude in fascicle lengthening (Finni, et al., 2003). Similarly, between methods of eccentric loading (isokinetic vs isotonic), there is no difference within the magnitude of fascicle lengthening – even though there were significant differences within the toque curves across the same ROM (Guilhem, et al., 2011). This information demonstrates that the velocity and application of load have a minimal impact upon the magnitude of fascicle lengthening, with the greatest change in FL occurring at the end ranges of motion as the majority of the lengthening is taken up by the elastic components within the early stages (Ando, et al., 2018).

Ando et al. (2018) demonstrated that the magnitude of lengthening is individual to single muscles, even within the same muscle group. The vastus intermedius underwent a significantly greater lengthening than the vastus lateralis, during the same eccentric task (Ando, et al., 2018). This further strengthens the thoughts that the potential of fascicle lengthening during active eccentric actions is related to individual tendon structural characteristics within muscles (Ando, et al., 2018). As the vastus lateralis possesses greater tendinous tissue than the vastus intermedius, resulting in a greater tendon lengthening in place of fascicle lengthening (Ando, et al., 2018). This observation would impact upon the exercise induced muscle damage of individual muscles, as the magnitude of fascicle lengthening is related to creatine kinase concentration, a key marker of exercise induced muscle damage (Hicks, Onambele-Pearson, Winwood, & Morse, 2017).

Within some muscle groups however, fascicle lengthening does not occur during eccentric conditions, within the tibialis anterior the muscle fascicles were observed acting quasi-isometrically (Reeves and Narici, 2003). The isometric action occurring at the fascicles coincided with a decreased PA, both of which acted independently of isokinetic velocity (Reeves and Narici, 2003). This highlights the role of the elastic component acting as mechanical buffer, by lengthening at the tendon to provide elastic potential energy – which would have important implications or energy saving during locomotion (Reeves and Narici, 2003). Although this was observed from a single image, at a single joint angle, whereas changes in FL may be observed if a continued series of measurements were made through the entire movement.

2.9.3.1 Dynamic actions within the hamstrings

A variety of single- and multi- joint resistance exercises have been studied previously using dynamic US, however, currently only one study has focussed upon the BF_{LH} (Cataneo, 2018), with much of the research focussed upon plantar flexors and knee extensors (Ando, et al., 2018; Ando et al., 2014; Cronin, et al., 2011; Earp, et al., 2017; Heroux, et al., 2016). The study of the structural behaviour of the muscle fascicles of the BF_{LH} was carried out by Cataneo (2018), where they used dynamic US to examine the dynamic structural changes of the BF_{LH} within hamstring rehabilitation exercises. The Askling protocol (Askling, et al., 2014), consists of three exercises, including: extender, glider and diver, that emphasise lengthening under tension, where Cataneo (2018) observed the greatest fascicle and MTU lengthening occurring within the glider (Figure 2-18), followed by the diver and extender exercises.

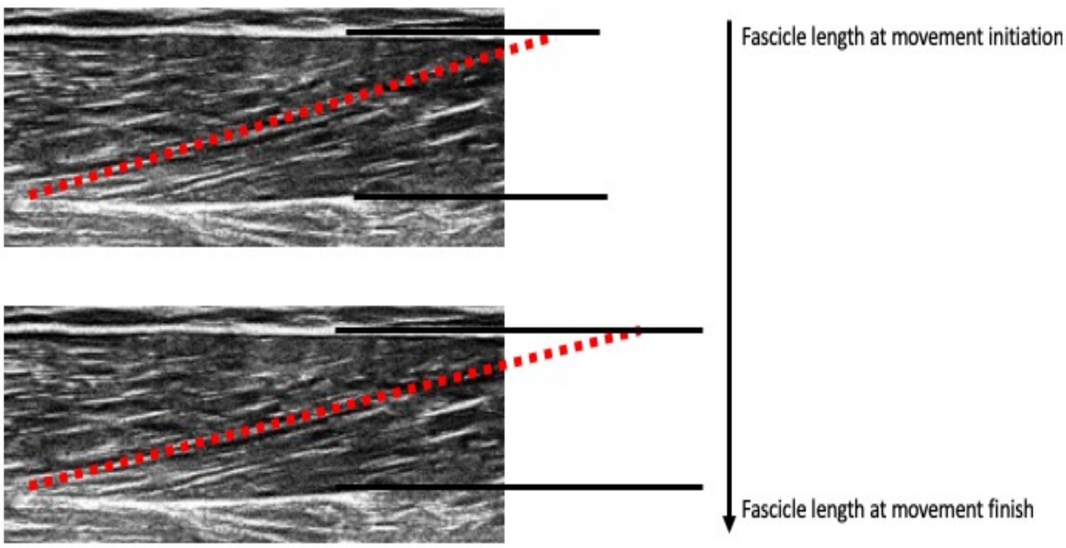


Figure 2-18 Representation of the 18% increase in BFLH FL during the glider.

Despite the authors providing novel information on the *in-vivo* dynamics of lengthening actions within the hamstrings, their observations contrast much of the previous literature that has distinguished the in-vivo dynamics eccentric actions of various muscles (Ando, et al., 2018; Ando, et al., 2014), where they typically incorporate both shortening and lengthening changes to the muscle fascicles. With both isometric and eccentric actions occurring, with simultaneous lengthening of the MTU, indicating greater utilization of the passive structures within eccentric muscle actions (Figure 2-19).

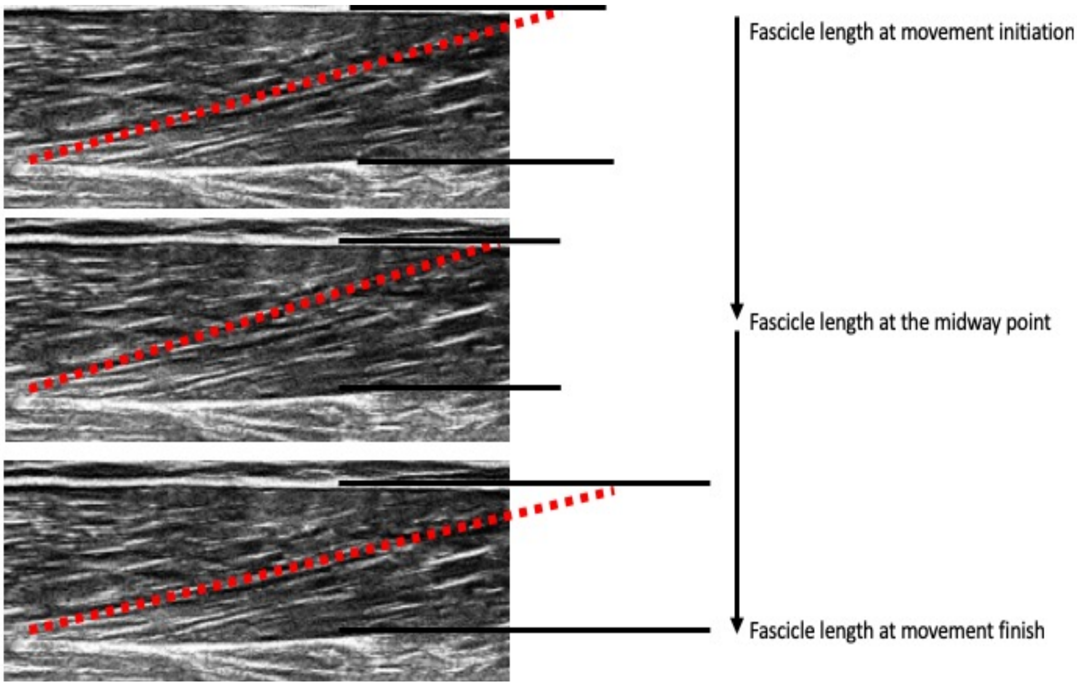


Figure 2-19 Hypothetical in-vivo functioning of the BFLH FL during a supramaximal eccentric exercise (i.e., nordic hamstring exercise).

The contrasting observation could be a question of hamstring loading as the Askling exercises are of low force, therefore there is minimal requirement of the muscle fascicles to actively shorten against the applied load (Askling *et al.*, 2014). However, the study performed by Cataneo (2018) is not without its limitations, as the researchers only observed the images at the beginning and end of the movement, therefore missing the dynamics involved during the exercises, thus making it difficult to make clear conclusions from this about typical behaviour of the muscle fascicles within the hamstring. During a rehabilitation phase of training, both high- and low-force exercises should typically be performed, commonly at low, controlled velocities with an emphasis on the lowering portion (Askling, et al., 2014). Therefore, as the Askling exercises are of low force, it could be of greater importance to observe higher intensity exercises, such as the NHE, which is a true supra-maximal eccentric exercise, in addition to the NHE being most effective exercise in reducing HSI rates (Bourne, et al., 2018; Shield and Bourne, 2018; Shield and Murphy, 2018), that we currently know of.

2.9.3.2 Methodological considerations

When attempting to perform dynamic US, there are several methodological considerations that need to be considered as potential risks of error that need to be mitigated. These include probe placement and orientation, FOV, frame rate and FL assessment (Van Hooren, Teratslas, & Hodson-Tole, 2020). These factors have been explored within a recent systematic review, where Van Hooren, Teratslas, et al. (2020) highlighted a number of methodological factors in great detail. However, primarily image quality is a key consideration prior to assessing dynamic fascicle changes of an investigated muscle under action – with the different hamstring muscles possessing varying planes and depths in active and passive conditions (Cataneo, 2018). This could be a result of the tuning fork shape the hamstrings possess (Balius, Pedret, Iriarte, Saiz, & Cerezal, 2019), with varying changes in plane and rotation under tension, in addition to a prominent muscle "bulge" that can also occur under tension. Therefore, an appropriately designed cast or external cuff is required to attach the probe to the area of interest where it is required to maintain appropriate contact with the skin in order to achieve the greatest image quality possible. Although some motion is inevitable while muscles undergo dynamic tasks (i.e., skin movement and subcutaneous tissue), minimising any potential motion is crucial in preventing erroneous results.

Availability of adequate US technology is another factor, particularly for high velocity movements (Van Hooren, Teratslas, et al., 2020). As task or movement speed increases, the in vivo muscle dynamic changes would also increase in velocity. Therefore, high speed US or higher frequency imaging should be considered to accurately assess the higher velocity dynamic muscle changes. Generally, within the literature sampling frequencies of less than 96 Hz has been utilised, with a range of between 25 – 200 Hz (Van Hooren, Teratslas, et al., 2020), which could be impacting upon the observed magnitude of change by under sampling during higher velocity tasks. Further research should look to identify the effect of sampling frequency on dynamic measurements, in both high- and low-velocity tasks, to identify a minimum required sampling frequency.

Once an acceptable image quality has been achieved, the analyses requires a method of identifying, measuring and tracking a fascicle of interest and its associated changes (Van Hooren, Teratslas, et al., 2020). This task is simple for muscle groups such as the medial gastrocnemius or soleus, where with even relatively small US transducers (e.g., < 6 cm) a compete fascicle could be in view (Van Hooren, Teratslas, et al., 2020). However, muscles such as the vastus lateralis and BF_{LH} (Van Hooren, Teratslas, et al., 2020), whereby fascicles can be in excess of 10 cm, it becomes difficult to track through an entire ROM, therefore estimation measures are required. Within the literature, even for static US image assessment, an array of methods has been utilised, with minimal agreement demonstrated between methods, with an approximate ~6% underestimation for the vastus lateralis (Van Hooren, Teratslas, et al., 2020). Furthermore, using a probe with a shorter FOV (e.g., < 6 cm), would exaggerate the differences observed between methods for muscles containing longer fascicles, such as the hamstrings (Franchi, Fitze, et al., 2019; Franchi, et al., 2018). Despite these analytic differences, researchers continually report absolute measures of FL change, whereas a recommendation for future research would be to report relative change or as a percentage of initial length and resting length (Van Hooren, Teratslas, et al., 2020). Measurements that were made as a relative percentage change would permit more appropriate comparisons to be made between individuals and tasks. This is particularly crucial for hamstring-based tasks, due the range of FLs identified and the effect this could have on the functioning characteristics.

2.10 Objectives of the Research

Within the literature review, a series of gaps were highlighted that require investigation within the research, including BF_{LF} FL, eccentric hamstring strength, kinematics of running with respect to low BF_{LF} FL and eccentric hamstring strength and hamstring exercise prescription, which has provided a series of objectives of the present thesis. The objectives will attempt to answer the overarching aim of observing HSI modifiable risk factors within system-based approach, which could be used within practice (i.e., measurement and identification, influence on performance and effect of training).

1) The assessment of BF_{LH} architecture using large single image FOV

Recently, in the assessment of BF_{LH} architecture, US methodologies have started to progress with research-based recommendations to move away from a single image estimation (Brennan, Cresswell, Farris, & Lichtwark, 2017; Franchi, Fitze, et al., 2019; Freitas, et al., 2018; Pimenta, et al., 2018), to more extensive methodologies such as linear and non-linear extended FOV - suggesting that these methods could provide greater accuracy over single image methods (Franchi, Fitze, et al., 2019; Pimenta, et al., 2018). However, the time available for practitioners to perform physical assessments has remained consistent (and limited), particularly at the elite senior levels. Therefore, extensive methods such as the extended FOV methods, could be impractical within elite practice, due to the time required to collect and analyse data. This is supported with literature, with even the most recent of intervention studies using a small FOV (< 6 cm) and single image estimations (Mendiguchia, et al., 2020), thus demonstrating the consistency of methods used since its inception into practice, despite the recent recommendations (Franchi, Fitze, et al., 2019; Franchi, et al., 2018). Consequently, the US methods that have been used previously and those that could be employed as availability of US technology improves such as larger single image FOV, requires exploration.

2) The role of eccentric hamstring strength and BF_{LH} FL upon kinematic and neuromuscular patterns during running.

The influence of a previous HSI upon the neuromuscular requirements across tasks (dynamic and static), as well alterations in biomechanical patterns has been previously explored (Fyfe, et al., 2013; Lord, Ma'ayah, et al., 2018; Navandar, Veiga, Torres, Chorro, & Navarro, 2017; Opar, Williams, et al., 2013b; Silder, Thelen, et al., 2010). However, there are some suggestions that these alterations could be part of a protective mechanism (Blandford, et al., 2018). In addition, despite a clear understanding of the modifiable risk factors of HSI (eccentric hamstring strength and BF_{LH} FL), to date there is limited exploration of how these risk factors could influence the performance of a high-risk task, such as running. Identifying potential changes in kinematics and neuromuscular requirements of running, as a result of reduced eccentric hamstring strength and BF_{LH} FL, could highlight potential mechanisms of HSIs during running based tasks, specifically pre-injury, which may offer some insight for a practitioner of what technical aspects should be observed during running.

3) The kinematic and activation patterns of the NHE and its variations.

Kinematic, neuromuscular and *in vivo* muscle mechanics (i.e., fascicle change) of exercise could aid researchers and practitioners in understanding potential adaptation mechanisms and aid in training prescription. The NHE is the most effective training modality in reducing HSI incidence (Chapter 2.5), it is also highly effective at increasing eccentric hamstring strength and BF_{LH} FL (Chapter 2.6,2.7). Despite this, many coaches and practitioners continue to disregard this information, opting to not perform the NHE (Bahr, et al., 2015; McCall, et al., 2016), although recent trends in practice have included assisted variants (Alt, et al., 2018; Matthews, Jones, Cohen, & Matthews, 2015). To date, two interventions studies have utilised an assisted variation of the NHE and have demonstrated positive changes in eccentric hamstring strength (Alt, et al., 2018; Matthews, et al., 2015). However, the methods applied are typically hard to standardise, therefore alternative options should be explored.

> 4) Explore the architectural and functional adaptations of the hamstrings using a mixed modal approach within an ecologically valid HSI prevention strategy.

In an effort to improve HSI prevention practices, there has been a recent push on research on sprint-based training as an effective modality (Freeman, et al., 2019; Mendiguchia, et al., 2020). However, the research is currently limited to only two vastly different intervention studies; with different prescriptions, dependent variables and populations (Freeman, et al., 2019; Mendiguchia, et al., 2020). Furthermore, intervention studies have typically embodied a single modality practice, i.e., only NHE, sprinting, razor curl, which 1) lacks the ecological validity of a complete training programme and 2) when considering the importance of HSI prevention practices a single modality approach would be unheard of, with a multi-modal hamstring approach common place including eccentric exercise, sprinting and hip dominant based training. Therefore, the effect of a multi-modal prescription requires investigation to determine the grouped effects on eccentric hamstring strength and BF_{LH} FL.

The major questions to be addressed in this programme of research include:

1) How does a large single image FOV US assessment influence the measurement made of the BF_{LH}?

2) Do athletes who possess high risk attributes (low eccentric hamstring strength and BF_{LH} FL) with no history of HSI, display altered kinematic and neuromuscular patterns during running?

3) What are the kinematic, neuromuscular, and *in vivo* muscle mechanics of the BF_{LH} during the NHE and its variations?

4) How does BF_{LH} FL and eccentric hamstring strength adapt to multiple modes of evidence-based training stimuli?

Within figure 2-20, the key questions which this thesis aimed to answer, with their associated studies have been presented.

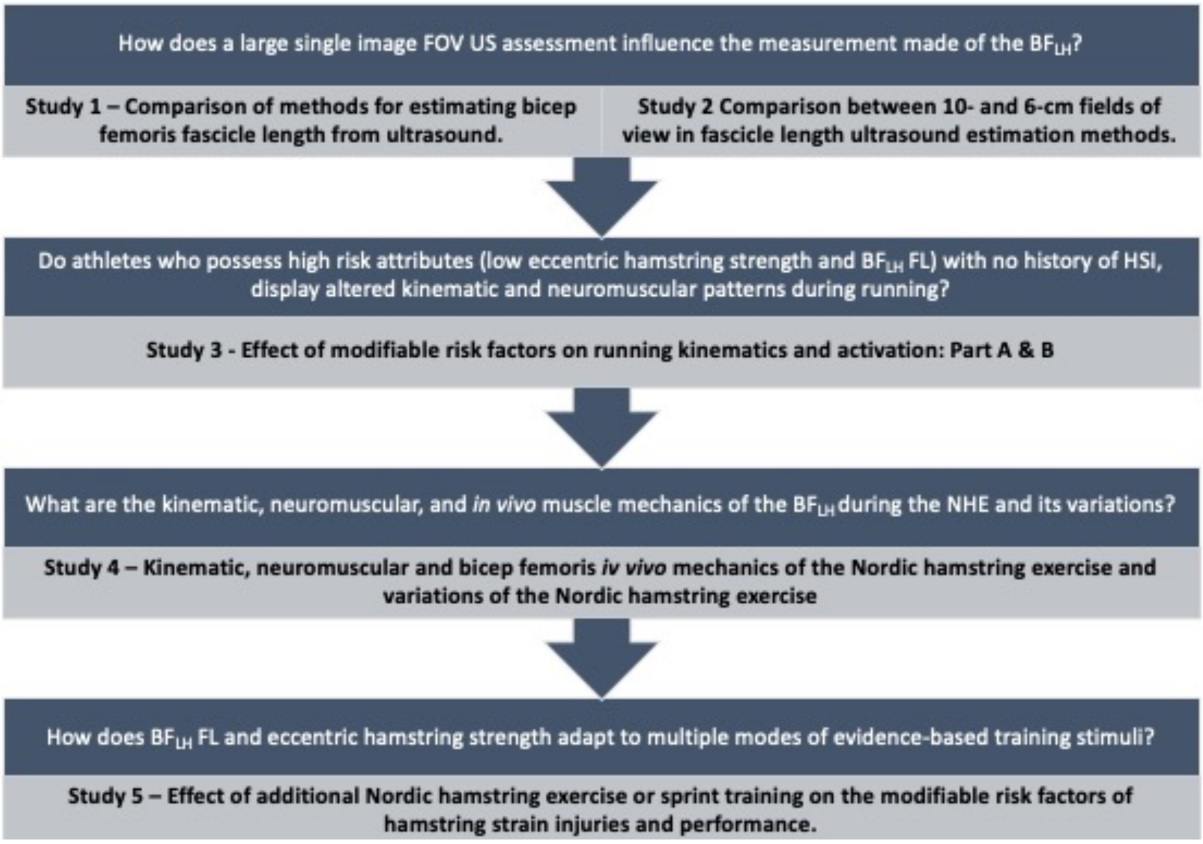


Figure 2-20 Schematic diagram identifying the major research questions and their associated studies.

3 Study 1 – Comparison of methods for estimating bicep femoris fascicle length from ultrasound.

3.1 Introduction

The complex architecture that makes up the BF_{LH} is potentially due to its diverse functioning, as a biarticular muscle with multiple roles in both injury prevention and athletic performance (Bourne, et al., 2018). In the role of injury prevention, FL of the BF_{LH} potentially may have the greatest influence, impacting upon the muscle's force-velocity and force-length relationships (Timmins, Shield, Williams, Lorenzen, et al., 2016). Due to the observed relationship between BF_{LH} FL and HSI incidence, measuring the BF_{LH} FL via the use of US has become common practice within elite sports, with sport specific normative values on BF_{LH} FL (Bourne, et al., 2018; Timmins, Bourne, et al., 2016). Within professional soccer, it has been reported that possessing a BF_{LH} FL of < 10.56 cm increases the risk of sustaining a HSI 4.1-fold (Bourne, et al., 2018; Timmins, Bourne, et al., 2016).

Currently within the research, using US images alone it is nearly impossible to completely measure the entire length of the BF_{LH} FL from a single image; as the FLs generally exceed the FOV of the probe (a typical probe length is 4-6 cm) (Abe, et al., 2016; De Oliveira, et al., 2016; Franchi, Fitze, et al., 2019; Franchi, et al., 2018; Freitas, et al., 2018; Pimenta, et al., 2018). As the whole fascicle is generally unviewable within a single US image, it has traditionally been estimated via a combination of tangible architectural measurements and trigonometry (De Oliveira, et al., 2016; Franchi, Fitze, et al., 2019; Freitas, et al., 2018; Kellis, et al., 2009; Lima, et al., 2015; Pimenta, et al., 2018). A criterion method of estimating FL (Equation 3.1) as proposed by Blazevich et al.(2006) and Kellis et al.(2009) includes measuring the AA or curvature of the fascicle in relation to the horizontal plane; in addition to the PA and MT and then proceeding to use trigonometry calculations to estimate FL. A secondary method presented within the literature, originally proposed for assessment of the vastus lateralis by Guilhem and colleagues (2011), was used recently by Freitas et al.(2018) and Pimenta et al.(2018) to estimate BF_{LH} FL. This involves partially measuring a visible fascicle and estimating the smallest portion which is not within the FOV (Equation 3.2). Previous research focusing on more symmetrical pennate muscle (vastus lateralis, triceps brachii) has utilised a more simplistic equation that does not take into account the AA (Equation 3.3) (Kawakami, et al., 1993). However, an enlarged FOV may reduce the influencing effect of the AA on the BF_{LH} FL - therefore reducing the complexity of the analysis, increasing time efficiency.

All methods of BF_{LH} FL estimation have been shown to be highly reliable and can be used to estimate BF_{LH} FL (Ando, et al., 2014; De Oliveira, et al., 2016; Franchi, Fitze, et al., 2019; Freitas, et al., 2018; Kellis, et al., 2009; Pimenta, et al., 2018; Timmins, et al., 2015). However, it would be hypothesised that methods which reduce the degree of estimation, via an increased single FOV, could potentially increase the accuracy and reliability of estimated measures and improve the agreement between FL estimation methods. To the authors knowledge, previous studies have compared FL estimation methods include US estimation versus cadaver specimens and a single image estimation versus an extended FOV image measurement (Franchi, Fitze, et al., 2019; Kellis, et al., 2009; Pimenta, et al., 2018). Studies have demonstrated that a single image trigonometry estimation, significantly overestimated

BF_{LH} FL compared to an extended FOV method (Franchi, Fitze, et al., 2019; Pimenta, et al., 2018), although this was not significantly different to a manual extrapolation performed upon a single image (Franchi, Fitze, et al., 2019). Single image estimations also demonstrated significant overestimations to cadaver specimens in BF_{LH} FL (Kellis, et al., 2009), in addition to large percentage differences (≥14.8%) from direct measurements (Ando, et al., 2014). However, no study to date has compared between the methods of estimating BF_{LH} FL when utilising a probe which enables an increased FOV. The purpose of this study, therefore, was to determine the reliability and to compare between the varying estimation methods when utilising a probe with a large FOV. It should be noted that this technique differs from previous US techniques that identify an extended FOV, as it is encompassed within a single US image captured by a large probe.

3.2 Methods

A test-retest design was used to assess BF_{LH} architectural parameters, including FL, across three equations derived from a large single probe with a FOV (10 cm). Six images of the BF_{LH} were captured with the 10 cm US probe across two-sessions (three per session) within a 7-day period for both the left and right legs. The researcher captured and digitized all images collected across both sessions. Between-session reliability was established across both time points. The study was approved by the institutional ethics committee (HSR1718-040) and conformed to the principles of the Declaration of Helsinki (1983).

3.2.1 Subjects

Thirteen physically active males (age 24.1 ± 3.8 years, body mass 79.3 ± 14 kg, height 179 ± 6.6 cm) with no history of lower-limb injury or inflammatory conditions completed two testing sessions. All participants were asked to refrain from any exercise 24 hours prior to each testing session. All participants reported that they participated in team sports on a regular basis (soccer = 6, rugby = 4, futsal = 2 and American football = 1). Written informed consent and the results of a health questionnaire (Appendix five), was obtained from all participants prior to testing.

3.2.2 Procedures

3.2.2.1 Ultrasound image acquisition

The scanning site for all images was determined as the halfway point between the ischial tuberosity and the lateral epicondyle, along the line of the BF. Images were recorded while participants lay relaxed in a prone position, with the hip in neutral and the knee fully extended. Images were subsequently collected along the longitudinal axis of the muscle belly utilizing a 2D, B-mode US machine (MyLab 70 XVision, Esaote, Genoa, Italy) with a 7.5 MHz, 10 cm linear array probe with a depth resolution of 67 mm.

To collect the US images, a layer of conductive gel was placed across the linear array probe; it was then placed on the skin over the scanning site and aligned longitudinally to the BF and perpendicular to the posterior thigh. During collection of the US images, care was taken to ensure minimal pressure was applied to the skin, as a larger application of pressure distort images leading to temporarily elongated muscle fascicles. The assessor (NJR) manipulated the orientation of the probe slightly if the superficial and intermediate aponeuroses were not parallel. These methods are similar to those used in earlier research (Timmins, Bourne, et al., 2016; Timmins, Ruddy, et al., 2016; Timmins, et al., 2015).

3.2.2.2 Architectural digitization

All sonograms were analysed off-line with Image J version 1.52 software (Wayne Rasband National Institute of Health, Bethesda, MD, USA). Images were first calibrated to the known length of the FOV, then for each image a fascicle of interest was identified, which was not always visible within the image. Finally, MT, PA, AA and observed FL were measured three times within each image, to enable complete FL estimation. Three trigonometric linear equations were utilised within the present study using the landmarks identified (equation 3-1, 3-2 & 3-3, Figure 2-14 & 2-15):

$$FL = sin(AA + 90deg)\ x\ MT/sin(180deg - (AA + 180deg - PA)$$

Equation 3-1 Criterion method of fascicle length estimation.

$$FL = MT/(sin\ (PA))$$

Equation 3-2 Fascicle length estimation using basic trigonometry

$$FL = L + (h\ \div\ sin\ (\beta))$$

Equation 3-3 Fascicle length estimation partial measure equation.

Where L is the observable fascicle length, h is the perpendicular distance between the superficial aponeurosis and the fascicles visible end point and β is the angle between the fascicle and the superficial aponeurosis.

3.2.3 Statistical analyses

Statistical analysis was performed using SPSS software version 25 (SPSS, Chicago, Illinois, USA) and Jamovi (Jamovi project (2018) Computer Software, retrieved from https://www.jamovi.org). A custom Microsoft Excel spreadsheet was also utilised. Statistical significance was set at *P* < 0.05 for all tests. Normality for all variables was confirmed using a Shapiro Wilks-test.

Within and between-session reliability based on the mean of each architectural parameter for each session, was assessed via a series of two-way mixed effects ICCs, 95% CIs and CV. A paired samples t-test and Cohen's *d* ES were utilized to determine if there were any significant differences between the session means. Minimum acceptable absolute reliability was confirmed using a CV <10% (Hopkins, 2000). The ICC values were interpreted based on the lower bound CI as (<0.50) poor, (0.5-0.74) moderate, (0.75-0.90) good and (>0.90) excellent (Koo and Li, 2016). SEM was calculated using the formula; $(SD(Pooled) \times (\sqrt{1} - ICC)$ (Thomas, Nelson, & Silverman, 2010), whereas the smallest detectable difference (SDD) was calculated from the formula; $\left((1.96\ \times \left(\sqrt{2}\right)\right) \times SEM)$ (Wright, 2002).

A repeated-measures analysis of variance (RMANOVA) and Bonferroni post hoc comparisons were conducted to identify if there were significant differences in the overall means FL values between the different estimation methods. Cohen's *d* ES and 95% CI were also calculated to determine the magnitude of differences. All Cohen's *d* ES were interpreted as trivial (<0.19), small (0.20-0.59) (Hopkins, 2002b).

The mean of difference between measures (bias) was expressed absolutely and as a percentage, ratio (criterion method/alternative method), 95% limits of agreement (LOA) (LOA: mean of the difference ± 1.96 standard deviations) and 95% CI were calculated between FL estimate methods using the methods described by Bland and Altman (Bland and Altman, 1986). The potential for hetero- or homoscedastic spread was assessed visually using the Bland and Altman plots. Unacceptable LOA were determined a priori as bias percentage greater than ±5% (Hansen, Cronin, & Newton, 2011). Pearson's correlation coefficients and coefficients of determination (R^2) were used to determine the relationship between the three FL estimation methods. Correlations were interpreted using the scale described Hopkins (2002b): trivial (0.0-0.1), small (0.1-0.3), moderate (0.3-0.5), large (0.5-0.7), very large (0.7-0.9), nearly perfect (0.9-1.0), perfect (1).

3.3 Results

Near perfect, between-session reliability was observed for all measures and estimations, with no significant difference between testing sessions. The mean values, reliability statistics, SEM, SDD and observed percentages for BF_{LH} architectural measurements are presented in Table 3-1. The average MT and PA was 2.70 ± 0.35 cm and $16.11 \pm 2.24°$ respectively.

Table 3-1 Between-session average, reliability and error statistics for bicep femoris long head architectural measurements.

	Muscle thickness (cm)		Pennation angle (°)		Criterion Measure (cm)		Basic Trigonometry (cm)		Partial Measure (cm)	
Within-session average	2.72		16.15		10.28		9.99		10.09	
Within-session SD	0.01		0.15		0.16		0.03		0.06	
Within-session CV (95% CI)	0.47 (0.46 – 0.48)		0.92 (0.86 – 0.97)		1.57 (1.51 – 1.64)		0.26 (0.25 – 0.27)		0.57 (0.55 – 0.60)	
Within-session ICC (95% CI)	0.951 (0.909 – 0.976)		0.937 (0.884 – 0.969)		0.892 (0.805 – 0.946)		0.940 (0.890 – 0.971)		0.987 (0.975 – 0.994)	
Test Occasion	**1**	**2**	**1**	**2**	**1**	**2**	**1**	**2**	**1**	**2**
Between-session average	2.72	2.69	16.15	16.07	10.28	10.32	9.99	9.94	10.09	10.13
Between-session SD	0.02		0.06		0.03		0.04		0.03	
Between-session CV (95% CI)	0.71 (0.70 - 0.72)		0.35 - (0.33 - 0.38)		0.35 (0.32 - 0.38)		0.37 (0.35 - 0.39)		0.32 (0.31 - 0.34)	
Between-session ICC (95% CI)	0.972 (0.939 - 0.987)		0.971 (0.937 - 0.995)		0.989 (0.972 - 0.995)		0.989 (0.975 - 0.995)		0.989 (0.985 - 0.999)	
p	0.11		0.45		0.52		0.37		0.60	
***d* (95% CI)**	0.09 (-0.63 - 0.46)		0.04 (-0.58 - 0.51)		0.02 (-0.52 - 0.56)		0.02 (-0.57 - 0.52)		0.02 (-0.52 - 0.57)	
SEM	0.06		0.38		0.28		0.21		0.24	
SEM%	2.17		2.36		2.71		2.11		2.37	
SDD	0.16		1.06		0.55		0.58		0.41	
SDD%	6.03		6.55		5.34		5.86		3.94	

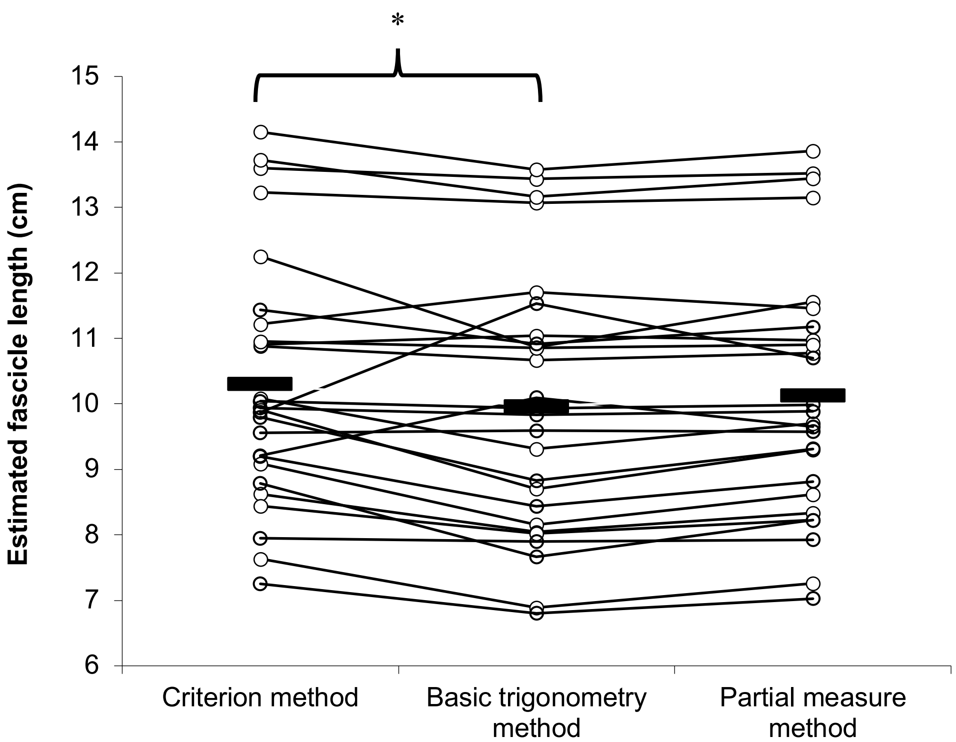


*Figure 3-1 Differences in estimated fascicle length between the three methods of estimation, * = significant difference (p < 0.05). Black line signifying mean estimated fascicle length, where circles signify individual measurements.*

The mean FL for each equation is displayed in Figure 3-1. The FL attained via the criterion method was significantly greater than the basic trigonometry method (p = 0.016), although this was only trivial in magnitude (d [95% CI] = 0.17 [-0.58 to 0.93]). Non-significant and trivial differences were observed between the criterion measure to the partial measure method (p = 0.081, d [95% CI] = 0.10 [-0.65 to 0.86]), and between the partial measure method and basic trigonometry method (p = 0.286, d [95% CI] = 0.08 [-0.68 to 0.84]).

Both the basic trigonometry and partial measure methods demonstrated unacceptable LOA (Table 3-2) (>5% (Hansen, et al., 2011)), when compared to the criterion measure. Individual Bland and Altman plots (Figure 3-2 A & B) illustrated heteroschedascity results between both methods in comparison to the criterion method.

Table 3-2 Bias and limits of agreement between the estimated measures of bicep femoris fascicle length.

			95% Limits of Agreement			
			Lower	**to**	**Upper**	**Ratio (SD)**
Criterion Vs Basic Trigonometry	**Bias (cm)**	0.334	-0.955	-	1.623	1.04 (0.05)
	95% CI	0.069 to 0.600	-1.415 to -0.495	-	1.163 to 2.083	
	Percent Bias (%)	3.24	-9.27	-	15.76	
Criterion Vs Partial Measure	**Bias (cm)**	0.188	-0.844	-	1.220	1.02 (0.04)
	95% CI	-0.025 to 0.401	-1.213 to 0.476	-	0.852 to 1.589	
	Percent Bias (%)	1.83	-9.19	-	11.84	

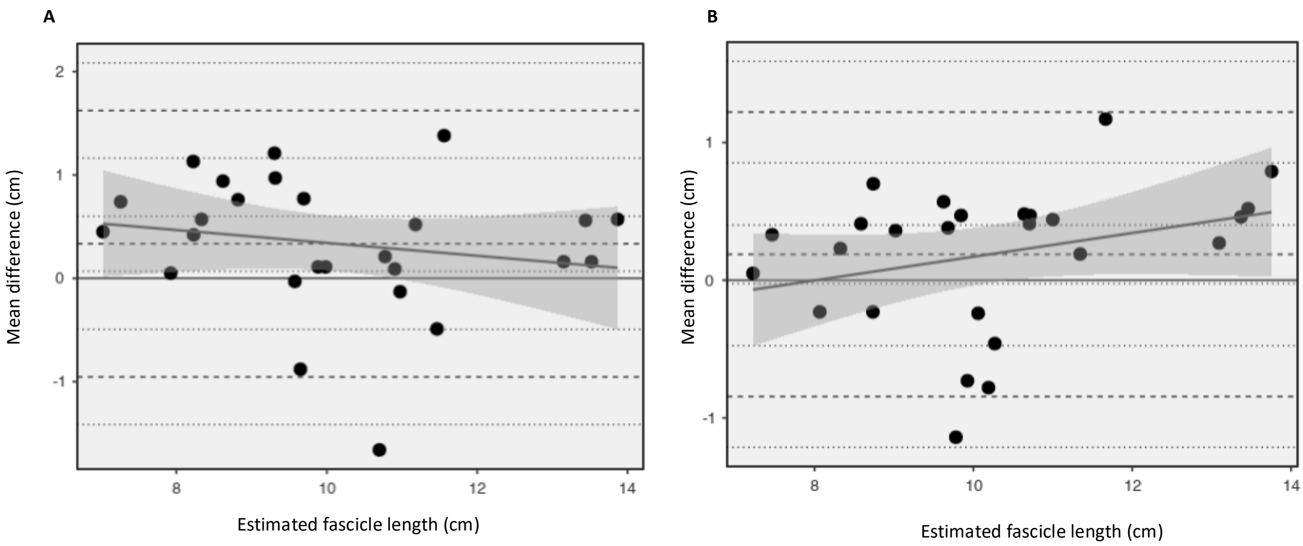


Figure 3-2 Bland Altman plots comparing the mean estimated fascicle lengths between methods. A) criterion vs. basic trigonometry and B) criterion vs. partial measure methods.

Almost perfect significant relationships were observed between the basic trigonometry and partial measure methods in comparison to the criterion estimation method (Table 6, Figure 3-3 & 3-4). Given the near-perfect relationships observed between the methods, linear-regression equations were established to allow FL estimations to be corrected between methods (Table 3-3).

Table 3-3 Observed relationships between the estimated measures of bicep femoris fascicle length

	Pearson's r (95% CI)	R^2	*p*
Criterion Vs Basic Trigonometry	0.945 (0.879 - 0.975)	0.893	< 0.001
Criterion Vs Partial Measure	0.961 (0.914 - 0.983)	0.924	< 0.001

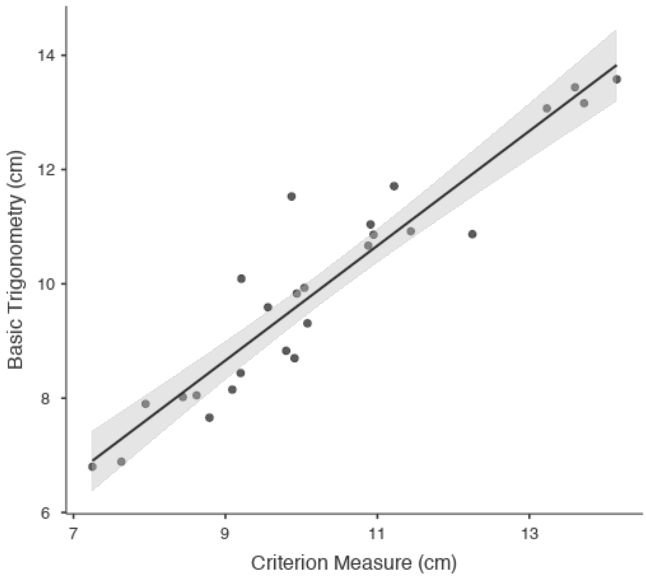


Figure 3-3 Relationship and 95% confidence limits between the criterion and basic trigonometry methods of estimating bicep femoris long head fascicle length.

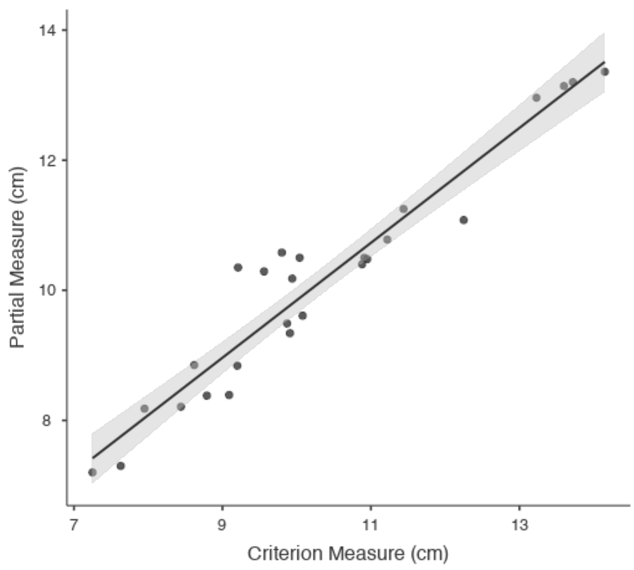


Figure 3-4 Relationship and 95% confidence limits between the criterion and partial measure methods of estimating bicep femoris long head fascicle length.

3.4 Discussion

The three estimation methods all reached minimum acceptable and near perfect between-session reliability (Table 3-1). The greatest relative reliability was observed for the partial measure method, whereas the greatest absolute reliability was seen with the criterion estimation method. A significant, albeit a trivial difference was observed between the criterion and basic trigonometry methods, whereas non-significant and trivial differences were observed between all other methods. Between the criterion and alternative methods an unacceptable degree of bias (>5%) was observed, with very large and near perfect relationships between the methods.

For the BF_{LH}, architectural measurements and the criterion method and partial measurement method have demonstrated high ICCs albeit slightly lower than the current study: 0.790 – 0.980, 0.800- 0.992 and 0.905- 0.960 for FL, MT and PA, respectively (De Oliveira, et al., 2016; Franchi, Fitze, et al., 2019; Freitas, et al., 2018; Kellis, et al., 2009; Pimenta, et al., 2018; Timmins, et al., 2015). The greater levels of reliability identified within the present study when compared to the previous research could be explained by a number of factors - firstly the inclusion of specific populations within previous research, women (De Oliveira, et al., 2016; Freitas, et al., 2018; Pimenta, et al., 2018), non-trained males (Ruas, Pinto, Lima, Costa, & Brown, 2017) and cadaver specimens (Kellis, et al., 2009), could have all impacted upon the US image quality, potentially by an increase in subcutaneous and intramuscular adipose tissue as well as effect of mortality on muscle characteristics, thus impacting on the reliability of measurements. This contrasts the participants within the present study who were all males and regular participants of team sports and resistance training. Secondly, the probe utilized within the present study had a FOV of 10 cm, this is in contrast to all previous work that has utilized shorter probes ~6 cm (De Oliveira, et al., 2016; Franchi, Fitze, et al., 2019; Freitas, et al., 2018; Kellis, et al., 2009; Pimenta, et al., 2018; Timmins, et al., 2015). This greater FOV could have aided in the accuracy and improved reliability of the measurements. Although the larger 10 cm probe has not been compared to its smaller counterparts within the literature, an extended FOV method using a single 6 cm probe to collect multiple images also demonstrated greater reliability and smaller SEM than a single image estimation (Pimenta, et al., 2018).

The partial measure method demonstrated the smallest SEM, followed by the criterion and the basic trigonometry method (Table 3-1). This trend is understandable, as the partial measure method estimates the smallest portion of the BF_{LH} FL, in comparison to both the criterion and basic trigonometry methods – as they both take an estimate of the entire FL of the BF_{LH}. The SEM identified within the present study are all comparable to those presented within previous research (De Oliveira, et al., 2016; Freitas, et al., 2018; Kellis, et al., 2009; Pimenta, et al., 2018; Ruas, et al., 2017; Timmins, et al., 2015). Timmins and colleagues (2015) presented SDD values for FL, MT and PA for US measurements for both left and right limbs independently, with SDD values for MT (0.16 cm and 0.22 cm, PA (1.02° and 0.88°) and FL (0.61 cm and 0.88 cm) identified, respectively (Timmins, et al., 2015). These results are similar to those presented within the present study utilising the criterion measure, as used by Timmins et al.(2015). The partial measure method does possess a lower SDD in comparison to the other methods, due to the greater observed reliability and smaller SEM. Both the present study and previous research indicate that meaningful changes over time could be identified using any FL estimations (Timmins, et al., 2015). Although, it should be noted that within the present study the muscle architecture data for each limb were pooled together, which is in contrast to Timmins et al.(2015) who observed limbs separately. It may be essential for practitioners to identify limb specific SDD values, as significant changes could be observed between limbs – potentially as a result of a previous HSI (Timmins, et al., 2017; Timmins, et al., 2015). Within the present study, this was not deemed essential analyses, as the current sample were not going be observed on multiple occasions, where alterations in FL from training or injury were going to be observed. The aim was to determine the difference between measurements and the researcher's ability to accurately collect and analyse US images over time.

Between the methods, there was minimal bias and with only trivial differences identified, with very large and near-perfect relationships between the criterion and alternate estimation methods. However, the individual Bland and Altman plots (Figure 3-2 A & B) illustrated heteroschedascity data between both methods in comparison to the criterion. This finding demonstrates that developing correction equations was unnecessary as they would provide a poor ability to correct any of the resultant values appropriately, therefore they were not developed. Additionally, it was found and proposed that there would be no fixed systematic error between the uncorrected or corrected estimations, with both under- and over-estimations identified across the mean bias. Highlighting, that despite agreement between methods, comparisons between should not be made.

Previously, it has been demonstrated that the partial measure method attained from a single image, significantly overestimated BF_{LH} FL in comparison to extended FOV methods – where the entire fascicle could be measured (Pimenta, et al., 2018). An extended FOV requires the ultra-sonographer to manually move the probe along the muscle according to the fascicle direction in either a linear or non-linear fashion to image an entire fascicle

(Pimenta, et al., 2018). A possible explanation for the significant difference between a single image vs extended FOV, could be the reduced accuracy of the single image by the use of the shorter 6 cm US probe (Pimenta, et al., 2018). In addition, the curvature of either the aponeurosis and or the fascicle are not considered within the partial measure method – the effect of which can be observed by the significant difference in fascicle angle measured from a single image compared to the linear extended FOV images (Pimenta, et al., 2018). The authors concluded that using different sonographic techniques (e.g., estimation measures, probe length), can affect the conclusions derived from the results due to the significant complex curvature to the BF_{LH} fascicles (Franchi, et al., 2018; Pimenta, et al., 2018).

3.5 Conclusion

In conclusion, the present study demonstrates that all methods of estimating BF_{LH} FL possess exceptionally high reliability. In addition, it also demonstrates that the assessor (NJR) has the ability to collect and analyse US images on multiple occasions, which would be crucial to be able to determine meaningful differences as an effect of a training intervention, when using the same US methodologies in future studies of this thesis. Although only trivial differences identified between methods, with minimal mean bias (<5%); the 95% LOA were unacceptable (>5%) indicating that the methods cannot be used interchangeably. Furthermore, developed correction equations ultimately did not improve the bias – increasing both the bias and the LOA; indicating it may not be appropriate to correct estimated FLs between methods.

With regards to recommendations for practitioners the commonly used phrase by researchers and practitioners alike - "it depends", is highly relevant. For accurate estimations, that limit the estimation process, the partial measure equation would be recommended. Although, if practitioners are limited by time, then the basic trigonometry could be valid option for practitioners, as this is the most simplistic equation requiring minimal measurements. However, there is limited normative data utilising these methods, in comparison to the criterion measure and requires further exploration, specifically with alternative FOV.

3.6 Linking paragraph

The current chapter demonstrated that the researcher (NJR) is highly reliable when using all of the BF_{LH} FL estimation methods when using a 10 cm FOV, providing SEM and SDD values which could be used to determine meaningful changes in the future from the effect of training or between groups. Chapter 3 also provided further insight into using alternative estimation equations to the criterion, with very strong and nearly perfect associations between equations. However, both alternative methods did not achieve acceptable LOA, with incongruent results and that despite the significant agreement between methods, corrected values were of an enlarged bias in the opposite direction with larger LOA. Despite the differences within the current study, it is the first study to utilise a 10 cm FOV to assess single image estimations. However, across the literature, all but one study has utilised FOV of <6 cm; therefore, the differences identified could be exaggerated a shorter FOV, as was found with extended FOV methods. It remains to be seen if the differences in FL estimations are influenced by FOV, therefore, comparisons of single image estimations should be performed across FOV which have been used within the literature and practice.

4 Study 2 Comparison between 10- and 6-cm fields of view in fascicle length ultrasound estimation methods

4.1 Introduction

The BF_{LH} is a biarticular muscle with multiple roles in both injury prevention and athletic performance (Koulouris and Connell, 2005; Lieber and Ward, 2011), functioning as both a hip extensor and knee flexor (Morin, et al., 2015; Schache, et al., 2013). The FL of the BF_{LH} has been reported to influence the muscle's force-velocity and force-length characteristics (Timmins, Shield, Williams, Lorenzen, et al., 2016). An increased FL through the addition of in-series sarcomeres, which results in a rightward shift of the force-velocity and force-length curves could be contributing to the observed relationship between absolute BF_{LH} FL and an elevated risk of HSI (Timmins, Bourne, et al., 2016; Timmins, et al., 2015). For instance, within professional soccer, it has been reported that possessing a BF_{LH} FL of < 10.56 cm increases the risk of sustaining a HSI 4.1-fold (Timmins, Bourne, et al., 2016). Therefore, measuring the BF_{LH} FL via the use of diagnostic US has become common practice within elite sport (Ribeiro Alvares et al., 2019; Timmins, Bourne, et al., 2016; Timmins, Shield, Williams, Lorenzen, et al., 2016).

Technology availability is a current limiting factor within US assessment, with typical probe lengths ranging between 4 - 6 cm (Behan, et al., 2018; De Oliveira, et al., 2016; Kellis, et al., 2009; Pimenta, et al., 2018; Timmins, Bourne, et al., 2016; Timmins, et al., 2015). Therefore, it is not possible to completely measure the entire length of the BF_{LH} FL from a single image (Franchi, Fitze, et al., 2019); as the FLs generally exceed the probes FOV. As the whole fascicle is generally not in view, common practice is to utilise tangible architectural measurements and trigonometry to estimate BF_{LH} FL. As with the previous chapter, the criterion method of estimating FL (Equation 3.1) as proposed by Blazevich et al. (2006) and Kellis et al. (2009), includes measuring the AA (curvature of the aponeurosis in relation to the horizontal plane); in addition to the PA and MT proceeding to use trigonometry calculations to estimate FL. A secondary method originally proposed for assessment of the vastus lateralis by Guilhem and colleagues (2011), which has been used more recently to estimate BF_{LH} FL (Franchi, Fitze, et al., 2019; Freitas, et al., 2018; Pimenta, et al., 2018), by partially measuring a visible fascicle and estimating the smallest portion not within the FOV (Equation 3.2). On more symmetrical pennate muscle (vastus lateralis, triceps brachii) a third, more simplistic equation that does not take into account the AA or any partial measure (Equation 3.3) has been used. However, it is hypothesized that methods which reduce the degree of estimation, via an increased single FOV or partial measure, could increase the accuracy and reliability of estimated measures.

Previous research has demonstrated that all methods of BF_{LH} FL estimation are highly reliable when using shorter FOV and can be used to routinely estimate BF_{LH} FL (Franchi, Fitze, et al., 2019; Timmins, Bourne, et al., 2016; Timmins, et al., 2015). However, all studies have also demonstrated that utilizing a single image estimation, significantly overestimated BF_{LH} FL (Franchi, Fitze, et al., 2019; Kellis, et al., 2009; Pimenta, et al., 2018). With large percentage differences (≥14.8%) from direct cadaver specimens (Kellis, et al., 2009), and approximately a 5 - 20% and over estimation bias between different methods of US image acquisition and estimation equation (Franchi, Fitze, et al., 2019; Pimenta, et al., 2018). However, no study to date has compared between single image estimations between two varying FOVs (i.e., 6- vs. 10 cm). Therefore, the purpose of this study, was to compare BF_{LH} FL estimations between the two FOVs, 6- vs. 10 cm. It was hypothesised that there would be significant and meaningful differences between the single image estimations from the two FOVs.

4.2 Method

4.2.1 Subjects

Sixteen male team sport athletes (age 24.1 ± 3.8 years, body mass 79.3 ± 14.0 kg, height 179 ± 6.6 cm) with no history of lower-limb injury or inflammatory conditions, had three images of the BF_{LH} captured for both the left and right legs with a 10 cm width US probe. All participants were also asked to refrain from any exercise 24 hours prior to each testing session. The researcher collected and digitized all images collected across both sessions. Written informed consent and the results of a health questionnaire (Appendix five) was obtained from all participants prior to testing. The study was approved by the institutional ethics committee (HSR1718-040) and conformed to the principles of the Declaration of Helsinki (1983).

4.2.2 Procedures

4.2.2.1 Bicep Femoris Ultrasound Acquisition

Initially the scanning site for all images was determined as the halfway point between the ischial tuberosity and the lateral epicondyle, along the line of the BF_{LH}. Images were recorded while participants lay relaxed in a prone position, with the hip in neutral and the knee fully extended. Images were subsequently collected along the longitudinal axis of the muscle belly utilizing a 2D, B-mode US (MyLab 70 XVision, Esaote, Genoa, Italy) with a 7.5 MHz, 10 cm linear array probe with a depth resolution of 67 mm.

To collect the US images, a layer of conductive gel was placed across the linear array probe; the probe was then placed on the skin over the scanning site and aligned longitudinally to the BF_{LH} and perpendicular to the posterior thigh. During collection of the US images, care was taken to ensure minimal pressure was applied to the skin, as a larger application of pressure can distort images, leading to temporarily elongated muscle fascicles. The assessor manipulated the orientation of the probe slightly if the superficial and intermediate aponeuroses were not parallel.

4.2.2.2 Bicep Femoris Architectural Digitization

All sonograms were analysed off-line with Image J version 1.52 software (Wayne Rasband National Institute of Health, Bethesda, MD, USA). Images were first calibrated to the known length of the FOV, then for 6 cm digitization all images were cropped by 4 cm from the distal portion of the image to attain a 6 cm FOV (Figure 4-1). The distal portion was cropped as within the 10 cm FOV, there was observed tapering of the aponeurosis across all subjects towards the muscle tendon unit junction.

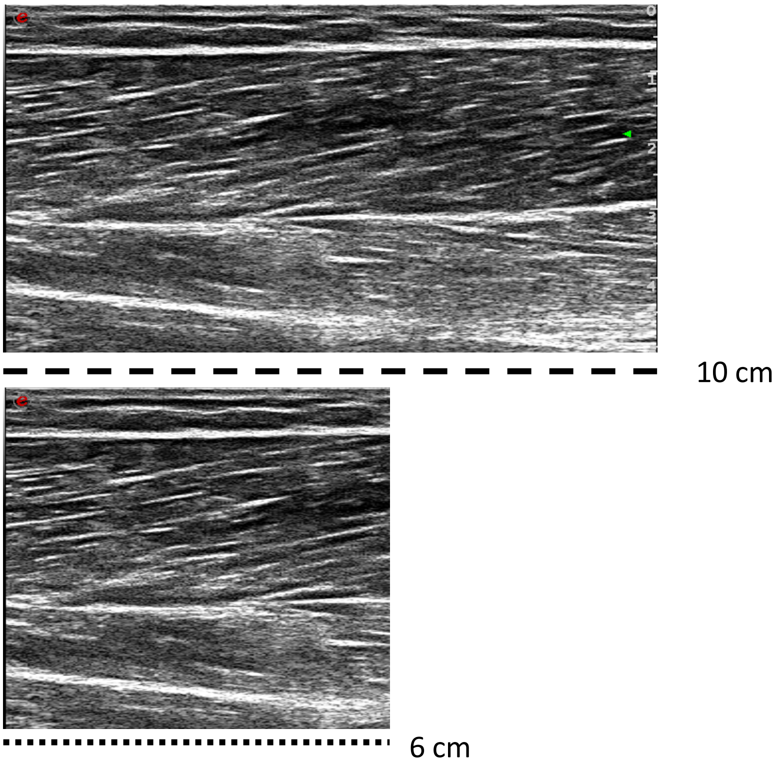


Figure 4-1 Representative scans from a single image ultrasound attained from a 10 cm FOV and a representative cropped 6 cm FOV, of the same scan and participant.

For each image (6- and 10 cm), a fascicle of interest was identified, where, MT, PA, AA and observed FL were measured three times within each image, to enable complete FL estimation. Three trigonometric linear equations were utilised within the present study:

$$FL = sin(AA + 90deg)\ x\ MT/sin(180deg - (AA + 180deg - PA)$$

Equation 4-1 Criterion method of fascicle length estimation.

$$FL = MT/(sin\ (PA))$$

Equation 4-2 Fascicle length estimation using basic trigonometry

$$FL = L + (h\ \div\ sin\ (\beta))$$

Equation 4-3 Fascicle length estimation partial measure equation.

Where L is the observable fascicle length, h is the perpendicular distance between the superficial aponeurosis and the fascicles visible end point and β is the angle between the fascicle and the superficial aponeurosis.

4.2.3 Statistical Analyses

Statistical analysis was performed using SPSS software version 25 (SPSS, Chicago, Illinois, USA) and Jamovi (Jamovi project (2018) Computer Software, Retrieved from https://www.jamovi.org). A custom Microsoft Excel spreadsheet was also utilised. Statistical significance was set at $P < 0.05$ for all tests. Normality for all variables was confirmed using a Shapiro Wilks-test.

Within-session reliability between the three collected images was assessed via a series of two-way mixed effects ICCs, 95% CIs and CV. Minimum acceptable absolute reliability was confirmed using a CV <10% (Hopkins, 2000). The ICC values were interpreted based on the lower bound CI as (<0.50) poor, (0.5-0.74) moderate, (0.75-0.90) good and (>0.90) excellent (Koo and Li, 2016).

Paired samples t-tests were conducted to determine if there were significant differences in the FL values between the different FOVs. Cohen's *d* ES and 95% CI were also calculated to determine the magnitude of differences using a custom excel spreadsheet. Cohen's *d* ES were interpreted as trivial (<0.19), small (0.20-0.59), moderate (0.60-1.19), large (1.20-1.99), and very large (≥2.0) (Hopkins, 2002a). Pearson's correlation coefficients, coefficient of determination (R^2) and explained percentage variance were used to determine the association between the FOV measures. Correlations were interpreted using the scale described Hopkins (2002a): trivial (<0.10), small (0.10-0.29), moderate (0.30-0.49), large (0.50-0.69), very large (0.7-0.89), nearly perfect (0.9-0.99), perfect (1).

The mean of the difference (bias) was expressed absolutely and as a percentage, ratio (criterion method/alternative method), 95% LOA (LOA: mean of the difference ± 1.96 standard deviations) and 95% CI were calculated between FOV measures using the methods described by Bland and Altman (Bland and Altman, 1986). The potential for hetero- or homoscedastic spread was assessed visually using the Bland and Altman plots. Unacceptable LOA were determined a priori as bias percentage greater than ±5%.

4.3 Results

All data was normally distributed ($p > 0.05$). With very high and nearly perfect, acceptable within session reliability for all methods of estimation for both 6- and 10 cm FOV methods (Table 4-1).

Table 4-1 Mean and standard deviation estimated bicep femoris fascicle lengths for 6- and 10 cm field of views and for each estimation method.

	6 cm Field of View			10 cm Field of View		
Estimation method	**Criterion**	**Basic Trigonometry**	**Partial measure**	**Criterion**	**Basic Trigonometry**	**Partial measure**
Mean (cm)	10.15	10.10	10.38	10.03	9.91	10.10
SD	1.52	1.53	1.58	1.73	1.78	1.55
CV	0.70	0.83	0.92	1.44	0.42	0.56
ICC (95% CI)	0.96 (0.92 - 0.98)	0.97 (0.95 - 0.99)	0.97 (0.93 - 0.98)	0.87 (0.76 - 0.94)	0.94 (0.88 - 0.97)	0.98 (0.97 - 0.99)

Non-significant, trivial to small differences were observed between 6- and 10 cm FOV methods for the criterion and basic trigonometry estimation methods, whereas a significant and small difference was observed between the partial measure estimation method. Between the FOV methods, significant ($p < 0.001$) very large and nearly perfect associations were observed (Table 4-2), with 74-81% of the estimated FL derived from the 6 cm FOV, able to explain the 10 cm estimations.

Table 4-2 Statistical differences and associations between the estimated bicep femoris fascicle lengths for 6- and 10 cm field of views for each estimation method.

	Differences		Association			
	p	***d (95% CI)***	***r* (95% CI)**	***p***	**R2**	**%**
Criterion 6 cm Vs 10 cm	0.420	0.136 (-0.178 - 0.416)	0.861 (0.743 - 0.927)	<0.001**	0.741	74.13
Basic Trigonometry 6 cm Vs 10 cm	0.137	0.254 (-0.065 - 0.453)	0.904 (0.819 - 0.950)	<0.001**	0.817	81.72
Partial measure 6 cm Vs 10 cm	0.049*	0.339 (0.000 - 0.546)	0.867 (0.753 - 0.930)	<0.001**	0.752	75.17
*** Denotes *p <0.05, ** denotes p < 0.001.***						

Unacceptable LOA (Table 4-3) (>5%) were observed, when the 6 cm FOV estimations compared to the 10 cm estimated FLs. Individual Bland and Altman plots and linear regressions (Figure 4-2), illustrating heteroscedastic results between 6- and 10 cm FOV estimations.

Table 4-3 Bias and LOA between the estimated measures of BF FL between 6- and 10 cm field of views.

			Limits of Agreement			Ratio (SD)
			Lower	**to**	**Upper**	
Criterion 6 cm Vs 10 cm	**Bias (cm)**	0.119	-1.601	-	1.840	1.03 (0.09)
	95% CI	-0.178 to - 0.416	-2.114 to -1.089	-	1.328 to 2352	
	Percent Bias (%)	1.18	-15.83	-	18.19	
Basic Trigonometry 6 cm Vs 10 cm	**Bias (cm)**	0.194	-1.307	-	1.695	1.02 (0.10)
	95% CI	-0.065 to 0.453	-1.754 to -0.860	-	1.248 to 2.142	
	Percent Bias (%)	1.96	-13.19	-	17.11	
Partial measure 6 cm Vs 10 cm	**Bias (cm)**	0.114	-0.907	-	0.854	1.03 (0.09)
	95% CI	0.008 to 0.546	-1.277 to -0.636	-	0.583 to 1.424	
	Percent Bias (%)	1.13	-8.98	-	8.45	

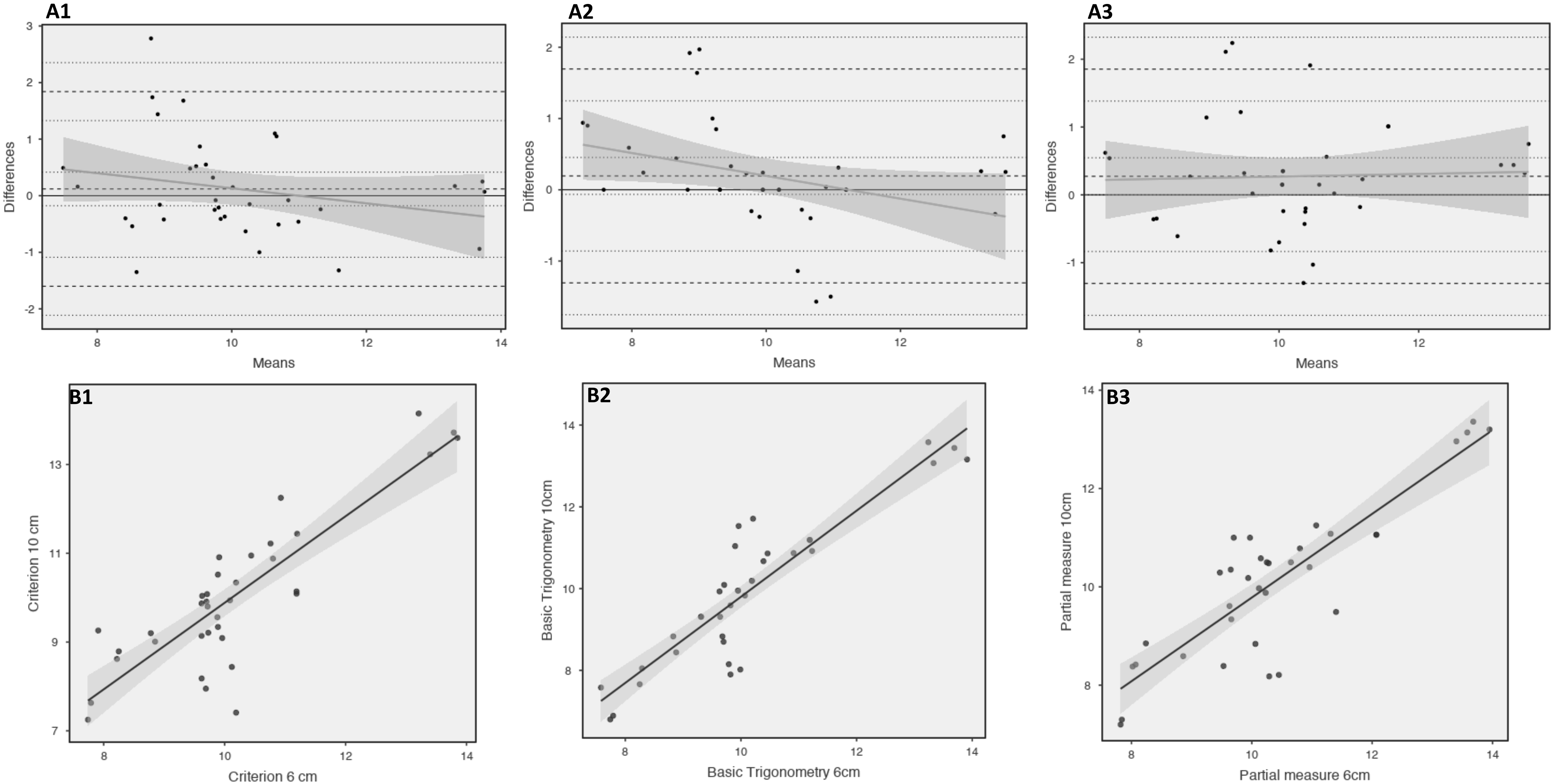


Figure 4-2 Agreement of estimated FL measurements between 6- and 10 cm FOV for each estimation equation A: Bland-Altman analyses showing absolute differences. B: Linear regressions for each estimation equation. 1: Criterion, 2: BT & 3: PM.

4.4 Discussion

The results of the present study demonstrated non-significant, trivial-small differences between 6- and 10 cm FOV, for the criterion and basic trigonometry estimation equations. Whereas a significant and small difference identified between 6- and 10 cm FOV, for the partial measure estimations. Furthermore, it is unadvisable to compare estimated BF_{LH} FL measurements between different FOV (6 and 10 cm) regardless of estimation equation used, with unacceptable LOA (>5%) found for all three estimation equations (Table 4-3). This is despite significant ($p < 0.001$), very large and nearly perfect associations observed between the different FOV (Table 4-2), where the 6 cm estimations were able to explain 74-81% of the 10 cm estimations.

One potential explanation for this significant difference, could be due to the reduced error within the partial measure equation when using the 10 cm FOV. When using the larger FOV (i.e., 10 cm), a greater degree of the observed fascicle can be measured when using the partial measure equation, thus reducing the error within the measurement. This could offer a potential explanation for the consistent overestimation when using the shorter 6 cm FOV, in comparison to the 10 cm FOV, which was not observed for the other estimation methods. Equation 4-2, is considered the most basic equation, using the fewest architectural characteristics in comparison to the alternative methods, this explains why 6 cm FOV was able to explain 81% of the variance within 10 cm estimations, as the tangible measurements were almost identical between FOV. Although, the remaining 19% could have as a result of differences in PA, as the angle of the fascicle relative to the deep aponeurosis could be observed more readily when utilising the larger 10 cm FOV.

The FOV length used to assess BF_{LH} muscle architecture, appears to be a crucial factor when using extrapolation methods to estimate BF_{LH} FL, including the estimation equations used within the present study. Research performed by Freitas et al.(2018), used methods which are consistent to those within the present study, whereby they observed two single image FOV, 3- vs. 6 cm, utilising the partial measure estimation equation to calculate FL. Although no comparative statistical analyses were performed on the differences between the calculated FLs, the 6 cm FOV estimated FLs were lower than the 3 cm comparison, 99.9 mm ± 15.7 and 120.3 mm ± 25.0 for the 6- and 3 cm FOV, respectively. This is consistent with the results of the present study, whereby the FL measurements achieved using larger FOV (10 cm) were smaller than those from the 6 cm FL, albeit only small and trivial differences. However, it would be expected that further reductions in FOV (< 6 cm) would lead to a greater magnitude in the difference between the estimations. This was highlighted by Franchi et al.(2019), who noted that the accuracy of extrapolation is dependent upon the length of the visible fascicle, with shorter FOV not permitting a large proportion of FL to be visually measured.

Despite the original values not reaching acceptable LOA (>5%), the criterion and basic trigonometry methods presented both under- and over estimations, suggesting that the accuracy of these estimation methods could be subject specific, with no fixed systematic error – which is consistent with previous findings (Franchi, Fitze, et al., 2019; Pimenta, et al., 2018). Conversely, the partial measure method of estimation presented a more fixed systematic error, with a FL overestimation when using the shorter FOV. Potentially indicating that the partial measure method of estimation could be used consistently between various FOV, as long as researchers and practitioners alike understand that any estimation values will likely be over-estimated when using shorter FOV (< 6 cm). Furthermore, the SEM values identified previously for the partial measure estimation equation for both the 6 cm and 10 cm FOV, are currently above the observed mean bias line and could therefore be erroneous. However, the LOA for the partial measure estimation equations were still unacceptable (<5%) although they were the best of the three chosen estimation methods.

Previously, single image estimations, such as the ones used within the present study, have been demonstrated to overestimate BF_{LH} FL and underestimated PA, in comparison to extended FOV methodologies, which aim to image the entire muscle in an attempt to attain an entire visible fascicle. A potential explanation of these differences includes the poor identification of the superficial aponeurosis trajectory, this is understandable as both studies that have compared static images to extended FOV have only used small FOV probes (5-6 cm), and not to a larger FOV as to which was used within the present study. Despite this, extended FOV methods requires complex algorithms and appropriate the application of technique (linear, non-linear or free hand), therefore are still not considered the "gold standard" for the assessment of BF_{LH} FL (Franchi, Fitze, et al., 2019; Franchi, et al., 2018). With a suggestion that research should be carried out to determine a "gold standard" method, with particular interest into the third dimension of the muscle – potentially minimizing the impact of fascicle curvature.

4.5 Conclusions

Within Chapter 3, it was suggested the partial measure equation was potentially the most accurate, as it removes the large degree of estimation found within the two other methods. However, as research still utilises single image estimations, potentially due to the reduced time required for the assessment and analysis in comparison to other recommended methods. The aim of the present study was to compare the FL estimations between the two FOVs, 6- vs. 10 cm. The results only partially met the hypothesis, with only one estimation equation presenting a small, but significant difference between FOV, with the reduced measurement error potentially leading to this difference, which was not observed between the other methods. This finding supports the recommendations from previous research suggesting that single image US estimations should be used with caution. However, advancements in US technology accessibility, including the availability of large FOV such as the 10-cm FOV used within the present thesis supports the suggestion, that a single image estimation could be appropriate for both research and practice. Despite the large associations found between the two FOVs they should not attempt to be compared, even when correction equations are applied – despite congruency between the equations the LOA were too large.

4.6 Linking paragraph

Chapters 3 and 4 have both explored the use of US to assess BF_{LH} muscle architecture, specifically BF_{LH} FL within research and practice. Shorter BF_{LH} FL has been proposed to be a modifiable risk factor in HSI occurrence, therefore, determining an optimal method of assessment was crucial to highlight who could be at an elevated risk. The criterion equation overestimates FL in comparison to the other equations, however, it is currently the only equation that has been used to highlight the risk of future HSI occurrence (Timmins, Bourne, et al., 2016). However, the findings of the two chapters highlighted that the partial measure equation could be optimal when a greater FOV is available, such as the 10-cm FOV. Eccentric hamstring strength is another modifiable risk factor for HSI, with slower isokinetic assessments (60 deg/s) having a small, yet predictive ability on subsequent HSI risk (Green, et al., 2018). Although the Nordbord has greater predictive validity to differentiate between those at risk of future HSI (Bourne, et al., 2015; Opar, et al., 2015; Timmins, Bourne, et al., 2016), the field-based device is currently not available. Furthermore, isokinetic dynamometry remains the gold-standard measure for single joint strength assessments. Both BF_{LH} FL and eccentric hamstring strength have a moderate body of evidence suggesting deficits in either, elevates the risk of HSI incidence within team sport athletes (Timmins, Bourne, et al., 2016). Research currently suggests that both of these factors could also play a key role in the mechanisms of HSI during running, as individuals who have suffered a previous HSI display differences in hamstring muscle activation, kinetic and kinematic profiles during running (Lee, et al., 2009; Lord, Ma'ayah, et al., 2018; Silder, Thelen, et al., 2010). Although these differences could be the effect of the injury, with potential muscular inhibition contributing to the observed differences (Blandford, et al., 2018). However, it is yet to be seen if the modifiable risk factors have an effect on running pre-injury, which could be an influencing factor on the mechanisms of HSI occurrence.

5 Study 3 - Effect of modifiable risk factors on running kinematics and activation: Part A & B

5.1 Part A – Effect of the modifiable risk factors of Hamstring strain injury upon Peak kinematic and activation during running

5.1.1 Introduction

The incidence of HSIs within sport are frequently result from performing one of two high risk actions, kicking or high-speed running (Opar, et al., 2012). The elevated risk of HSI occurrence posed by high speed running is due to the hamstrings required to produce extremely high forces, up to 10.5 N/kg for the BF_{LH} during the terminal swing phase (Nagano, et al., 2015), in order to resist rapid knee extension (Chumanov, et al., 2011; Navandar, et al., 2017; Opar, et al., 2012; Thelen, Chumanov, Hoerth, et al., 2005). Two case study have provided circumstantial evidence, identifying the approximate time of HSI event within treadmill running (Heiderscheit, et al., 2005; Schache, et al., 2009). Heiderscheit and colleagues (2005) approximated that a HSI event occurred at some point during the late swing phase or the very initial stance phases. Although it is impossible to determine exactly where the HSI occurred between the terminal swing phase or early stance phase, it could be proposed that it occurs during the terminal swing phase with it only being pronounced via the occurrence of pain during the subsequent foot contact, with the earliest indication of an injury occurring only 0.1 s following foot contact (Heiderscheit, et al., 2005; Schache, et al., 2009).

Currently within the literature, it is unclear as to what muscle action occurs during the terminal swing phase with some conflict reported within the literature. Generally, an eccentric muscle action is proposed to be occurring during the terminal swing phase (Heiderscheit, et al., 2005; Higashihara, et al., 2015, 2016, 2018; Schache, et al., 2012; Schache, et al., 2013; Schache, et al., 2009), using data derived from muscle simulation models and MTU length estimations, where there is lengthening across both the knee and hip joints (Heiderscheit, et al., 2005; Higashihara, et al., 2015, 2016, 2018; Schache, et al., 2012; Schache, et al., 2013; Schache, et al., 2009). This is contested within the literature with Van Hooren and Bosch (2017a) explaining that an isometric muscle action occurs during the terminal swing phase, further describing that if an eccentric action was to occur it could be the cause of HSI (Van Hooren and Bosch, 2017a, 2017b, 2018). However, this is purely conjecture, using animal models and citing that all current research defining an eccentric muscle action uses theoretical muscular modelling (Van Hooren and Bosch, 2017a, 2017b, 2018).

Regardless of what is happening at the muscle level, it has clearly been shown that hamstring strength is crucial in preventing HSI (Bourne, et al., 2015; Opar, et al., 2015; Opar, et al., 2012; Ruddy, Shield, et al., 2018; Timmins, Bourne, et al., 2016). Hamstring strength training interventions have indicated that the hamstring muscles are extremely pliable and adapt rapidly to the stimulus applied, specifically with the inclusion of an eccentric training stimulus, where there is a rapid increase in both hamstring strength and BF_{LH} FL (Bourne, et al., 2018). Both of these adaptive responses to eccentric hamstring focused strength training have not only demonstrated reductions in HSI occurrence (Askling, et al., 2003a; van der Horst, et al., 2015; van Dyk, et al., 2019), but also subsequent increases in performance of athletic tasks such as running and jumping (Chu, Yaremko, & VonGaza, 2017; Freeman, et al., 2019; Ishoi, et al., 2018; Krommes et al., 2017). This demonstrates that the performance of athletic tasks (e.g., running and jumping) may be associated with changes in hamstring strength and BF_{LH} architecture.

Individuals with impaired functioning, through either a history of HSI occurrence or acute fatigue, have demonstrated alterations in running kinematics, kicking mechanics, muscle activation patterns and lengthening muscle tissue mechanics (Brughelli, Kinsella, & Nosaka, 2011; Emami, Massoud Arab, & Ghamkhar, 2014; Lee, et al., 2009; Lord, Blazevich, et al., 2018; Navandar, et al., 2017; Silder, Thelen, et al., 2010; Small, et al., 2009). During submaximal running, researchers demonstrated that a previous HSI significantly reduced horizontal force production during high speed running (80% max velocity) (Brughelli, et al., 2011), potentially resulting in a decrease in peak hip flexion and the peak knee extensor moment that occurred during the late swing phase of running, although these differences were not assessed. Additionally, under fatigued conditions, where it would be expected that the ability for the hamstrings to produce force would be impaired, significant differences were highlighted in the swing phase kinematics in semi-professional soccer players (Small, et al., 2009). Contrastingly, however, Silder et al.(2010) demonstrated no significant differences in mechanics or muscular activation between previously and non-previously injured limbs when running at 60-, 80-, 90- and 100% of maximum sprinting speed.

Despite these studies providing detailed information around the functioning characteristics of the hamstrings during athletic tasks. They only observed the impact of a HSI event on hamstring and athletic performance retrospectively (Brughelli, et al., 2011; Emami, et al., 2014; Lee, et al., 2009; Lord, Blazevich, et al., 2018; Navandar, et al., 2017; Silder, Thelen, et al., 2010), whereby the kinetic, kinematic and activation changes could be a result of motor adaptation, where there is a change in technique or performance too optimize performance and/or protect the system from further injury (Hodges and Tucker, 2011). More recently, significant relationships have been identified between eccentric hamstring strength and late swing phase mechanics at the knee (Alt, et al., 2020). Furthermore, following a four-week NHE training programme, improvements in the late swing phase knee mechanics were also observed by the same research group (Alt et al., 2021). However, as they only observed the late swing phase, there could be meaningful changes in the individuals sprint technique or performance, prior to the late swing, therefore the observations of the whole gait cycle and EMG of the hamstrings is warranted. Furthermore, BF_{LH} FL is also a key factor in optimal functioning of the hamstrings during exercise therefore, further investigation into the these contributing factors is required. It could be of great benefit to practitioners to determine if differences in kinematic and muscle activation patterns are present during running in non-previously injured subjects who are both eccentrically weaker and possess shorter BF_{LH} fascicles. This information may identify what if any, meaningful changes may be present during running at difference speeds within individuals who are at a perceived higher risk of future HSI event.

5.1.1.1 Aims and Hypothesis

The primary aim of this study was to observe if any difference in running kinematics or hamstring activation patterning were present between individuals who possessed high or low eccentric hamstring strength and BF_{LH} FL. A secondary aim was to observe if any relationship existed between the measures of eccentric hamstring strength and BF_{LH} architecture.

It was hypothesised that a significant difference in running kinematics and muscle activation would exist between high and low risk groups. Additionally, strong associations would be observed between the modifiable risk factors for injury (i.e., hamstring strength and BF_{LH} architecture).

5.1.2 Methods

5.1.2.1 Participant characteristics

Eighteen physically active males (age 24.7 ± 4.3 years, height 181.9 ± 7.2 cm, mass 84.9 ± 12.9 kg) volunteered to participate in this study. There were no inclusion-exclusion criteria incorporated for this study. All participants performed competitive sport on a weekly basis, and they also incorporated high-speed running within their training collected via a questionnaire. All participants were of good overall health based on the completion of a Health Questionnaire (Appendix five) and had not suffered a previous HSI in the last 3 years. All participants reported that they performed regular team sport (collegiate – semi-professional), including football, rugby, American football, basketball, futsal, hockey, and lacrosse. Overall, none of the participants reported that they had previously performed any structured sprint training, with minimal exposure to technical elements being described during sport-based warm-ups.

The study was approved by the institutional ethics committee (HSR1718-040), and all participants had both read a Participant Information Sheet and provided written informed consent prior to testing. The study also conformed to the principles of the Declaration of Helsinki (1983).

5.1.2.2 Research Design

This study was completed based on an observational research design. Participants attended the Human Performance Laboratory at the University of Salford for testing on two separate occasions within a one-week period, each interspersed by ≥ 48 hours, at the same time of day.

Participants' modifiable risk factors were assessed during the first testing occasion whereby BF_{LH} muscle architecture was assessed via resting ultrasonography, following which they performed isokinetic strength measurements for the quadriceps and hamstrings in both concentric and eccentric modes of action. On the second testing bout, participants performed a submaximal treadmill assessment running at several speeds with both 3D and EMG measurements taken.

5.1.2.3 Protocol

5.1.2.4 Standardised warm up

Prior to the isokinetic strength assessments participants performed a standardised warm up which was performed following the collection of US images, which consisted of 5 mins of submaximal cycling, followed by two sets of five repetitions of body weight squats, lunges and leg swings. On the second testing occasion, prior to all running trials, participants performed a second standardised warm up, identical to the previous test occasion, with the addition of three submaximal 10 m skips and accelerations. A standardised warm up has been shown to be crucial for the collection of EMG data, with significant differences being identified in RMS amplitude during an isometric action between a warm up and a no-warm up control group (Stewart, Macaluso, & De Vito, 2003).

5.1.2.5 Data collection

5.1.2.6 Modifiable risk factor assessment

5.1.2.7 Muscle architecture

A full description of how US images of the BF_{LH} were collected can be found in Chapters 3 and 4. During US assessments the distance between the ischial tuberosity and lateral epicondyle was measured as an indication of BF_{LH} MTU length, which were subsequently utilised to determine FL relative to MTU length.

5.1.2.8 Isokinetic strength

Peak absolute and relative knee flexion and extension torque was assessed using an isokinetic dynamometer (125AP, KinCom, TN, USA) sampling at 120 Hz. Relative torque was deemed to be more appropriate for the present study, as allometrically scaling as a proportion of fat free mass would make no difference during running, as individuals will still have to decelerate the shank regardless of local (i.e., shank) composition during running. Additionally, at the time of assessment the University of Salford had no access to more accurate methods of body composition assessment, such as MRI. After the standardised warm up was completed, participants were seated on the dynamometer so that the hip was flexed to 90°, ensuring that the dynamometer lever arm and the knee joint centre were aligned (Figure 5-1). The trunk, waist and tested thigh were fixed with straps to minimise secondary joint movement. The ROM of the knee was determined as 0 to 90° (i.e., full extension to 90° of flexion), the limb length and limb weight for each participant was recorded, with limb weight being measured at rest at 0°, for gravitational correction during data analysis.

Figure 5-1 Image demonstrates the isokinetic dynamometry set up; current position represents measurement of limb weight at approximately full knee extension.

Testing was comprised of two different modes (concentric/concentric and eccentric/eccentric) involving the quadriceps and hamstrings at a single standardised angular velocity (60°/s). This angular velocity was chosen as a small predictive effect may be possible at this angular velocity to detect future HSI risk (Green, et al., 2018). Participants performed five submaximal incremental repetitions of knee extension and flexion prior to the performance of maximal efforts across both modes, this was used for familiarisation purposes. Following the familiarization, participants performed three maximal concentric knee extension and concentric knee flexion efforts. This was followed by three maximal eccentric knee extension and eccentric knee flexion efforts. A 60 s rest period was observed between each set. The assessor (NJR) provided instructions to either "push" or "pull" the dynamometer head as "hard and fast as possible", while additionally providing vigorous verbal encouragement, during all testing sessions. All raw torque/angle data received through the dynamometer was saved and exported as an ASCII file for later analysis using Shelton Technical Data Transfer software (Shelton Technical Limited, Milton Keynes, UK).

5.1.2.9 Running assessment

5.1.2.10 Motion capture

For each running trial, 3D motion data was collected over a 15-second duration using 10 Qualisys Oqus 7 infrared cameras (250 Hz) operating through Qualisys Track Manager Software (Oqus 7+, Qualisys AB, Partille, Sweden). Prior to commencing each testing occasion, the Oqus camera system required calibration to define the capture volume of the testing area. The calibration process is considered integral to ensuring the collection of valid three-dimensional motion data (Chiari, Croce, Leardini, & Cappozzo, 2005). Calibrating the Qualisys camera system required a two-step process, firstly, an "L" shaped frame containing four passive retro-reflective markers (each separated by a known distance) was placed on the lower right-hand corner of the final of 3 force platforms (Figure 17), ensuring that all four markers were visible to each camera. It should be noted that forces were not collected within the present study as there would be a large degree of interference through the treadmill. The position of the "L" shaped frame served as a reference to the position of the global coordinate system, ensuring that each infrared camera was able to identify where the defined origin of the human performance laboratory was (e.g., x-y-z = 0-0-0). The global coordinate system was defined with x representing the medio-lateral direction, y representing the anterior-posterior direction and z representing the vertical direction.

Secondly, a "T" shaped wand (with one passive retro-reflective marker placed at either end for a known distance) was moved around the testing area in all three orthogonal planes, whilst being recorded by the Qualisys camera system over a 60-second period. Following the calibration period Qualisys track manager software produced a calibration result as the average residual, which was used to determine whether the calibration was successful. A higher average residual is indicative of a poorer ability of the infrared cameras to measure the known distance between the markers located on the "T" shaped wand, whilst it was moved around the testing area. The Qualisys track manager manufacturer's guidelines recommend that the average residual should be ≤ 2.0 mm, however, calibration was considered successful when the average residual was ≤ 0.8 mm in the present study.

On completion of a successful calibration of the Qualisys camera system, several passive retro-reflective markers (14 mm in diameter) were placed onto the body landmarks of the legs and pelvis; to ensure marker placement consistency the same researcher performed this task. Six markers were placed upon the anterior superior iliac spine (x2), posterior superior iliac spine (x2) and the iliac crest (x2) defined the pelvis segment. The thigh segment of each leg was defined by three anatomical markers in total placed on the medial (x1) and lateral (x1) femoral epicondyles and the greater trochanter (x1) and a cluster set of four tracking markers attached to a lightweight rigid plastic shell secured to the anterior aspect of the thigh (at mid-length) with elasticated bandages. The shank segment of each leg was defined by two anatomical markers in total placed on the medial (x1) and lateral (x1) malleoli and a cluster set secured to the anterior aspect of the shank (at mid-length) with elasticated bandages. Each foot segment was defined by four anatomical markers in total placed on the first (x1), third (x1) and fifth (x1) metatarsals and the calcaneus (x1) (Jones, Herrington and Graham-Smith, 2016; Jones, Donelon and Dos'Santos, 2017), using standardised laboratory footwear. This was followed by a static trial where the participants stood within the testing area, with their lower limbs and pelvis orientated in a neutral anatomical position.

The treadmill (T9450HRT Vision Fitness, Cottage Grove, WI, USA) used for all submaximal running trials was situated within the 10-camera analysis area (Figure 6-2). The tester checked passive retro-reflective marker visibility manually during periods of rest between the running trials to ensure that a minimum of three markers per segment were visible always of the test.

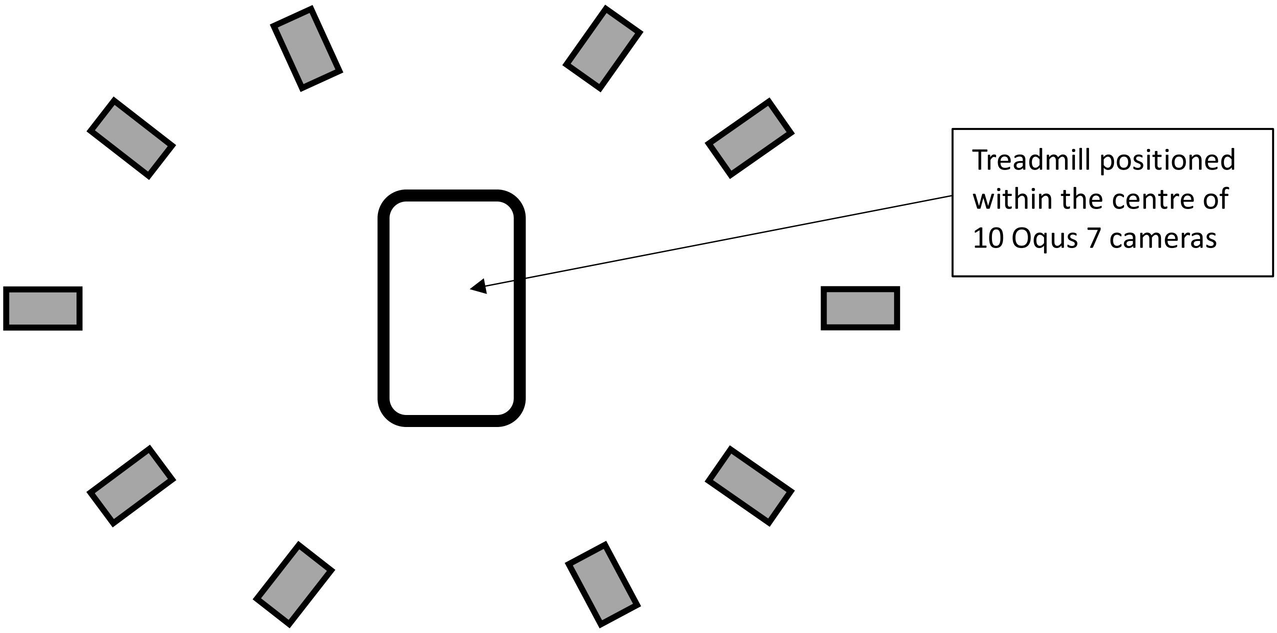


Figure 5-2 Visual representation of the treadmill positioning within the motion capture area.

5.1.2.11 Electromyography

5.1.2.12 Task electromyography

Surface EMG activity of BF_{LH} and ST was measured for all trials. Prior to electrode placement, the participants' skin was prepared using a standardised process of shaving (with a disposable safe razor), rubbing with a preparation gel and cleaning with an alcohol-based solution. Skin preparation was performed to minimise resistance (i.e., to reduce inter-electrode resistance to values below 5 kΩ (Ishikawa, Komi, Grey, Lepola, & Bruggemann, 2005).

A pair of self-adhesive Ag-AgCl electrodes ~10 mm in diameter (Noraxon Dual EMG electrode, Noraxon U.S.A Inc, Scottsdale AZ, USA) were placed on to the surface of the skin of both limbs. Electrodes were placed at the mid-point of the BF, identified via a line measured between the ischial tuberosity and lateral epicondyle. With the mid-point of the ST, identified via a line between the ischial tuberosity and the medial epicondyle. They were attached parallel with the orientation of the muscle fibres and in a bipolar configuration, with a constant inter-electrode distance of 17.5 mm**.**

Electrodes were attached to wireless EMG sensors via electrode leads, weighing <14 g (2B EMG Sensor, Noraxon U.S.A Inc, Scottsdale AZ, USA). Secured to the leg via double-sided adhesive tape, whilst confirming that no tape residue blocked the reference pad, which may interfere with the raw data. A reference pad was also applied, away from the placement of the electrodes for stable EMG readings. The wireless EMG sensor sent live information to a receiver (Desktop DTS Receiver, Noraxon U.S.A Inc, Scottsdale AZ, USA) connected to desktop computer.

Raw EMG data was captured at 1500 Hz, allowing for synchronization with 3D motion data within the Qualisys track manager software (C-motion, version 3.90.21, Gothenburg, Sweden). Correct electrode placement was confirmed prior to commencing data collection with manual muscle testing (i.e., by asking the participants to voluntarily contract the hamstrings against manual resistance) and minimal cross-talk will be visually and physically checked via internal and external rotation of the leg with a 90° knee angle, as per Timmins et al.(2014).

5.1.2.13 Normalization electromyography

A max-effort sprint assessment performed to normalize rectified task EMG data. This method was chosen to allow for comparisons of EMG attained at the sub-maximal running speeds to be expressed as a percentage of maximal velocity. After all sub-maximal treadmill trials were completed, a 4-minute rest period was provided during which all 3D markers, with exception of the thigh cluster sets, were removed to ensure a maximal effort with little interference of maximal running gait. Furthermore, the EMG electrodes used during all trials were securely located beneath the bandages concurrently holding the thigh cluster sets in position; therefore, both remained in place for maximal treadmill sprint testing.

Following the allocated 5-minute rest period, participants moved onto a different treadmill (Woodway Ergo ELG55, Weil am Rhein, Germany) that can attain higher running velocities, which was in a fixed position outside the 3D motion capture area. Participants then ran at increasing velocities where they were required to maintain a set running velocity for 10 seconds with 180 seconds recovery between each rep. Commencing at 18 km·hr^{-1}, with subsequent increases in velocity of 1.5 km·hr^{-1} for each running interval, similar to methodologies described by Numella et al.(2007). They continued this until they could not maintain the pace for the given duration or a rating of perceived exertion (RPE) of >9 was given, when using a scale of 1-10.

5.1.2.14 Data analysis

5.1.2.15 Ultrasound Analysis

A full description of how US images of the BF_{LH} were analysed can be found in Chapters 3 and 4. Briefly, the partial measure estimation equation was utilised (Equation 2-5), as Chapters 3 & 4 demonstrated the potential greater reliability and accuracy.

5.1.2.16 Isokinetic Analysis

Once all torque/angle data had been exported, it was analysed using a custom designed Excel spreadsheet. Phases of acceleration and deceleration were initially deleted from the analysis using a tolerance of ± 1°·s^{-1}. Following this, data was subjected to a gravitational correction process, where a gravitational correction value of the lower limb weight was calculated via multiplying the moment arm of force application by the limb weight using the following equation.

$$Gravitational\ Correction\ value = (Sin(Angle)\ x\ Lever\ Arm)\ x\ limb\ weight$$

Equation 5-1 Gravitational correction equation for isokinetic knee flexor/extensor assessment.

The correction value was subsequently summated or subtracted to all torque data dependent on which muscle group was being assessed. A correction value was added to all data points for Quadriceps during knee extension, due to the negative effect of gravity. Whereas they were subtracted for hamstrings during knee flexion, due to the positive effect of gravity.

Following the gravity correction, absolute and relative to bodyweight eccentric hamstring peak torque and concentric quadriceps peak torque values were identified for each trial, along with the corresponding angles at which the peak torque was achieved. To identify possible muscular imbalances (quadriceps vs hamstrings), the functional hamstring to quadriceps ratio was calculated as a measure of muscular asymmetry using the following equation.

$$Functional\ H\!:\!Q = Eccentric\ Hamstring\ Peak\ torque \div Concentric\ Quadriceps\ Peak\ Torque$$

Equation 5-2 Isokinetic functional hamstring to quadriceps ratio

Due to the associated criticisms of the dynamic ratios (Graham-Smith, et al., 2013; Green, et al., 2018), a further calculation of muscular asymmetry was calculated, namely the angle of crossover through dynamic control profiling. To obtain the dynamic control profile and derive the angle of crossover for each data point (angle), the torque angles for the concentric quadriceps were subtracted from the eccentric hamstrings. The point where the net joint torque crossed zero on the x-axis is the angle of crossover (Alt, Knicker, & Strueder, 2017; Graham-Smith, et al., 2013), where the greater the angle of crossover, the greater the ROM within which the hamstrings can eccentrically counteract the concentric action of the quadriceps (Alt, et al., 2017; Graham-Smith, et al., 2013).

5.1.2.17 Modifiable risk factor grouping

Participants were rank ordered and a qualitative score (i.e., 1 – 18) was applied to each measurement. The middle two subjects were subsequently removed to clearly define two groups (i.e., high and low risk). Due to the observed relationships between the relative measures of strength and FL, a further grouping was calculated using relative measures for both Part A and Part B.

5.1.2.18 Three-Dimension Motion analysis

All motion data were trimmed, and passive markers were labelled using QTM software (Qualisys AB, Partille, Sweden). The duration of each five second static trial was trimmed to approximately one second. The duration of each 15 second running trial was reduced to the time period between three successive strides within the gait cycle. All markers (i.e., anatomical and tracking markers) were individually labelled so that they could be accurately identified during the subsequent model building process. The static trial for each participant was manually labelled by the tester.

Following this (to improve the efficiency of the labelling process), an Automatic Identification of Markers model was created for each participant based on the labels allocated for the static trial and subsequently applied to the running trials (i.e., across all running velocities). The accuracy of the Automatic Identification of Markers models was individually checked by the tester and any inaccuracies were corrected by relabelling any mislabelled markers before saving the files. Upon completion of the trimming and labelling process in QTM, all successful trials were exported as individual C3D signal files (which included motion and time data) for further analysis.

A lower extremity six degrees of freedom kinematic model was created for each participant from the static trial. This included the pelvis, thighs, shanks and feet using Visual 3D software (C-motion, version 3.90.21, Gothenburg, Sweden). The local coordinate system was defined at the proximal joint centre for each segment, and all measurements were related back to the static trial or anatomical zero alignment. The model utilized a CODA pelvis orientation to define the location of the hip joint centre (Bell, Brand, & Pedersen, 1989). The knee and ankle joint centres were defined as the mid-point of the line between lateral and medial markers.

Normal methods of determining gait characteristics typically involve the use of vertical ground contact forces measured using imbedded force platforms. Therefore, to identify gait characteristics within the current study with the interaction of the treadmill, a novel method was required. A method was devised whereby left and right foot, take off (TO) and touch down (TD) events were identified within Visual 3D software (C-motion, version 3.90.21, Gothenburg, Sweden), using event threshold and onset pipelines functions. The minimum height of the fifth metatarsal when positioned on the treadmill was a measured height of 0.22 m from the ground, therefore, TO was identified as the moment the fifth metatarsal ascended (Z) to a height greater than this minimum threshold for a minimum of 25 frames (0.1 s). Alternatively, TD was identified via an onset pipeline function, as the moment the fifth metatarsal reached 0.22 m for a minimum of eight frames (0.032 s). Following the identification of gait events, data was stride normalized from TD to subsequent TD, with contact time being defined as TD to TO.

Peak hip and knee angles during the gait (Figure 5-3), were determined using an X-Y-Z Cardan sequence of rotations, which reflects the default joint coordinate system set in Visual3D and is determined from the previously mentioned local coordinate system for each rigid segment (Cole, Nigg, Ronsky, & Yeadon, 1993). In other words, sagittal plane knee and hip joint angles was determined based on the three-dimensional coordinates of one rigid segment relative to another (i.e., a reference segment). For example, the sagittal plane knee angle was determined based on the local coordinate system of the shank segment relative to those of the thigh segment, with the latter acting as the reference segment. Prior to exporting the first derivative kinematic angular data, an 8 Hz low pass filter was applied to the data to attenuate noise (Winter, Sidwall, & Hobson, 1974). During pilot data analysis of variety of filtering frequencies (no filter, 2-, 4-,6-, 8-, 10- and 12 Hz) were used and an 8 Hz filter was found to be optimal as it was able to attenuate noises within the signal, without impacting upon the true signal in comparison to both smaller and higher frequencies.

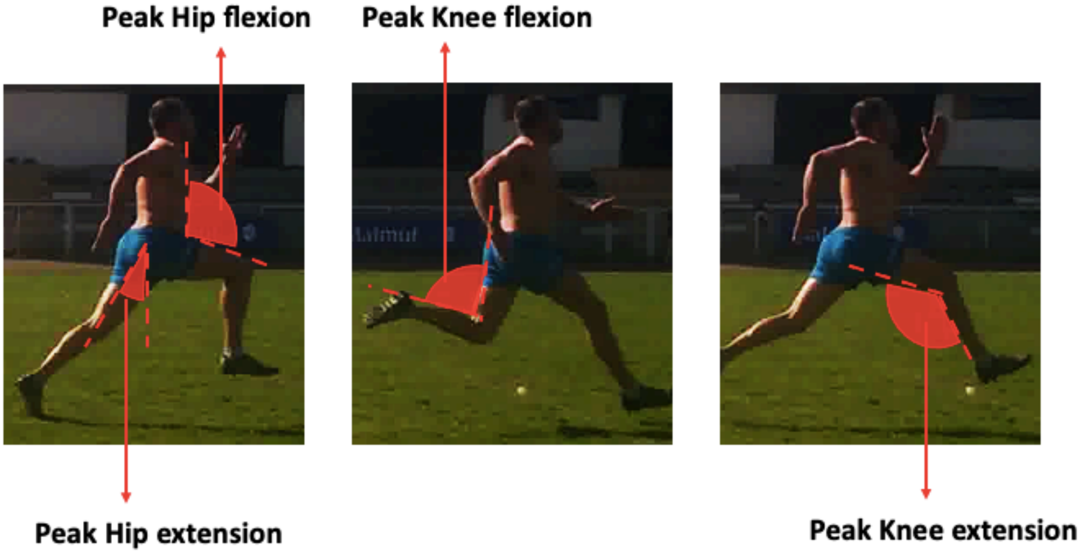


Figure 5-3 Visual representation of the peak hip and knee angles determined through the gait cycle.

The kinematic markers (Figure 5-3) were chosen as they represent key markers across the gait cycle that could aid in identifying lower limb mechanics to the knee and hip, additionally enabling the understanding any pelvic shift that might be occurring. Furthermore, they are the two joint angles which the hamstrings cross over, specifically the BF_{LH}, which are required to estimate BF_{LH} MTU length (Hawkins and Hull, 1990). Additionally, they are similar markers to those used within 2D analysis within the Altis Kinogram (Josse, 2020; McMillan and Pfaff, 2018) or the proposed 'kick-back' mechanism (Lahti et al., 2020).

5.1.2.19 Muscle-Tendon modelling

Estimations of BF_{LH} MTU lengths across the running gait were calculated using regression equations. Hawkins and Hull (1990) have identified constant values and algebraic equations that can estimate MTU lengths for all the muscles of the lower limb, including the BF_{LH} .

For the BF, Hawkins and Hull (1990) identified a quadratic regression equation of;

$$L = C0 + C1\chi + C2\beta + C3\beta^2 + C4\emptyset$$

Equation 5-3 Quadratic regression equation to estimate BF_{LH} FL (Hawkins and Hull, 1990).

Where L represents the normalized muscle length, C0-C4 the constant coefficients and χ, β and $\emptyset$ the hip, knee and ankle angles respectively. The BF_{LH} constant values are, 1.048, 2.09E-3, -1.60E-3, 0 and 0 for C0, C1, C2, C3 and C4 respectively (Hawkins and Hull, 1990).

5.1.2.20 Electromyography analysis

The raw EMG signals were initially high- and low-pass filtered between 10 and 1000 Hz, as pre-set filters within the Noraxon receiver (Desktop DTS Receiver, Noraxon U.S.A Inc, Scottsdale AZ, USA), before being exported from the Qualsys track manager software (C-motion, version 3.90.21, Gothenburg, Sweden). This type of initial data processing is performed to remove unwanted artefacts (including; noise artefacts (e.g. movement of the cables) and the identification of cardiac signal amplitude) (Gerdle, et al., 1999). The data was then exported into a custom Excel spreadsheet where further processing and analyses was performed. Within the custom Excel spreadsheet, processing of the EMG data continued with an RMS filter, across a moving average window of 25 ms. This filtering window was chosen as it presented high acceptable reliability for peak and mean EMG values for the BF during the glute-ham raise (GHR). Peak EMG amplitude of both the BF_{LH} and ST were identified across the normalized stride, in addition, a peak ratio of the BF_{LH} to ST was identified.

5.1.2.21 Statistical analysis

Statistical analysis was performed using SPSS software version 25 (SPSS, Chicago, Illinois, USA) and Jamovi (Jamovi project (2018) Computer Software, Retrieved from https://www.jamovi.org). A custom Microsoft Excel spreadsheet was also utilised. Statistical significance was set at $P < 0.05$ for all tests. Normality for all variables was confirmed using a Shapiro Wilks-test. Data are presented as mean ± SD. Absolute and relative between-trial and between-stride reliability was assessed by CV percentages and a two-way random effects model ICC, with 95% CI determined for both measures of reliability. Minimum acceptable reliability was confirmed using an CV <10% (Hopkins, 2000). The ICC values were interpreted based on the lower bound CI as (<0.50) poor, (0.5-0.74) moderate, (0.75-0.90) good and (>0.90) excellent (Koo and Li, 2016).

5.1.2.22 Modifiable risk factors statistical approach

Pearson's product moment correlation coefficients with 95% CIs and coefficient of determination (R^2) were used to determine if any relationships exist among isokinetic muscular qualities of the knee extensors and flexors and BF_{LH} muscle architecture based on absolute and relative data, for the pooled (left and right leg) data (n = 36). Correlations were interpreted using the scale described Hopkins (2002b), trivial (0.0-0.1), small (0.1-0.3), moderate (0.3-0.5), large (0.5-0.7), very large (0.7-0.9), nearly perfect (0.9-1.0), perfect (1).

Subjects were ranked ordered based on the eccentric hamstring strength and BF_{LH} FL, hamstring isokinetic strength characteristics (i.e. absolute peak eccentric torque, relative peak eccentric torque, angle of crossover and functional H:Q), BF_{LH} FL and as a combination of both strength and FL. The two middle subjects were then removed to form to separate groups. Dividing the subjects in this manner resulted in the high and low risk groups. Mean differences in each variable derived for high and low risk groups were compared using independent t-tests. Cohen's *d* ES were calculated to provide a measure of magnitude of the differences in each variable noted between groups and they were interpreted in line with previous recommendations, which defined values of < 0.35, 0.35-0.80, 0.80-1.5 and > 1.5 as trivial, small, moderate, and large respectively (Rhea, 2004).

5.1.2.23 Kinematic and electromyographic statistical approach

To observe the mains effect of increasing running velocity on mean normalized and peak kinematic and EMG variables, i.e., estimated BF_{LH} MTU length, peak knee flexion, peak hip flexion, peak BF_{LH} amplitude, peak ST amplitude, peak BF:ST amplitude ratio, a number repeated-measures analysis of variance (ANOVA) were performed. Mean ± SD peak kinematic and EMG data will be presented for high and low risk groups, as a function of the modifiable risk factors at each running velocity. To determine the difference between peak variables, a two-way ANOVA was performed across each running velocity with Bonferroni post-hoc corrections to observe the effect of the modifiable risk factor on each kinematic and EMG variable at each speed.

5.1.3 Results

All data was determined as being normally distributed (p >0.05).

5.1.3.1 Muscle architecture

Participants had BF_{LH} MTU lengths of 45.8 ± 4.9 cm and 45.7 ± 5.0 cm, for the left and right legs respectively at rest. Tables 5-1 and 5-2 show the repeatability outcomes of the BF_{LH} architecture parameters. For both left and right limbs, very high reliability (i.e., ICC >0.8, CV <10%) was observed for MT, PA and FL.

Table 5-1 Between image repeatability muscle architecture measurements for the right bicep femoris long head

	Muscle Thickness (cm)	Pennation Angle (°)	Fascicle Length (cm)	Relative Fascicle length
Image 1	2.77	16.27	10.04	0.22
Image 2	2.79	16.55	10.00	0.22
Image 3	2.77	16.15	10.27	0.22
Mean	2.78	16.33	10.10	0.00
SD	0.01	0.21	0.15	1.47
CV (95% CI)	0.36 (0.17-1.30)	1.27 (0.25-2.28)	1.47 (0.42-2.51)	1.59 (0.45-2.51)
ICC (95% CI)	0.97 (0.94-0.98)	0.94 (0.87-0.97)	0.89 (0.81-0.96)	0.89 (0.81-0.96)

Table 5-2 Between image repeatability muscle architecture measurements for the left bicep femoris long head

	Muscle Thickness (cm)	Pennation Angle (°)	Fascicle Length (cm)	Relative Fascicle Length
Image 1	2.72	16.27	9.74	0.21
Image 2	2.69	16.09	10.11	0.22
Image 3	2.68	15.88	9.95	0.22
Mean	2.70	16.08	9.94	0.22
SD	0.02	0.20	0.19	0.00
CV (95% CI)	0.73 (0.22-1.69)	1.21 (0.21-2.22)	1.88 (0.77-2.99)	1.88 (0.77-2.99)
ICC (95% CI)	0.97 (0.93-0.99)	0.95 (0.90-0.98)	0.88 (0.81-0.95)	0.88 (0.81-0.95)

5.1.3.2 Isokinetic strength

Tables 5-3 demonstrates the repeatability outcomes of the isokinetic strength measurements. For both left and right limbs, very high reliability (i.e., ICC >0.8, CV <10%) was observed for hamstring and quadriceps peak eccentric hamstring and concentric quadriceps absolute and relative torques, functional H:Q ratio as a function of absolute and relative measures and angle of crossover.

Table 5-3 Between trial repeatability for hamstrings and quadriceps isokinetic strength and asymmetry measurements.

Measurement		Trial 1	Trial 2	Trial 3	Average	SD	CV (95% CI)	ICC (95% CI)
Hamstrings Eccentric Action	Absolute Peak Torque Right (N·m)	148.11	148.28	145.39	147.26	1.62	1.10 (0.11-2.09)	0.96 (0.92-0.99)
	Absolute Peak Torque Left (N·m)	136.22	140.06	138.50	138.26	1.93	1.39 (0.36-2.42)	0.96 (0.91-0.98)
	Relative Peak Torque Right (N·m/kg)	1.60	1.60	1.57	1.59	0.02	1.10 (0.11-2.09)	0.96 (0.92-0.99)
	Relative Peak Torque Left (N·m/kg)	1.47	1.52	1.50	1.50	0.02	1.39 (0.36-2.42)	0.96 (0.91-0.98)
Quadriceps Concentric Action	Absolute Peak Torque Right (N·m)	214.56	213.28	214.06	213.96	0.64	0.30 (0.03-1.23)	0.98 (0.96-0.99)
	Absolute Peak Torque Left (N·m)	208.67	208.06	205.61	207.44	1.62	0.78 (0.18-1.74)	0.98 (0.96-0.99)
	Relative Peak Torque Right (N·m/kg)	2.84	2.89	2.81	2.85	0.04	1.31 (0.29-2.32)	0.98 (0.95-0.99)
	Relative Peak Torque Left (N·m/kg)	2.48	2.48	2.45	2.47	0.02	0.71 (0.25-1.66)	0.98 (0.96-0.99)
Functional Ratio H:Q	Absolute Peak Torque Right	0.72	0.73	0.71	0.72	0.01	1.49 (0.44-2.53)	0.95 (0.90-0.98)
	Absolute Peak Torque Left	0.67	0.69	0.68	0.68	0.01	1.44 (0.41-2.48)	0.97 (0.93-0.99)
	Relative Peak Torque Right	0.64	0.68	0.66	0.66	0.02	2.65 (1.39-3.92)	0.86 (0.72-0.94)
	Relative Peak Torque Left	0.60	0.63	0.62	0.62	0.01	2.39 (1.14-3.04)	0.91 (0.82-0.96)
Angle of Crossover Right (°)		26.39	25.36	24.83	25.53	0.79	3.10 (1.73-4.47)	0.91 (0.81-0.96)
Angle of Crossover Left (°)		23.44	24.58	22.83	23.62	0.89	3.76 (2.22-5.30)	0.95 (0.90-0.98)

5.1.3.3 Between risk factor associations

Pearson's correlations and coefficients of determination between eccentric hamstring strength and BF_{LH} FL are presented in Table 5-4. Specifically, absolute BF_{LH} FL demonstrated only small or trivial correlations with all isokinetic parameters of the knee flexors (Table 5-4).

Table 5-4 Relationship between isokinetic strength characteristics of the knee extensors and flexors and bicep femoris long head fascicle length

	Absolute Fascicle Length				Relative Fascicle Length			
	r (95% CI)	*p*	R^2 (%)	Correlation Coefficient Descriptor	*r*	*p*	R^2 (%)	Correlation Coefficient Descriptor
Absolute Hamstring eccentric Peak Torque	-0.089 (-0.406-0.246)	0.604	0.00 (0.78)	Trivial	0.379 (0.058-0.629)	0.023	0.14 (14.36)	Moderate
Relative Hamstring eccentric Peak Torque	0.284 (-0.49-0.56)	0.093	0.08 (8.07)	Small	0.920 (0.848-0.959)	<0.001	0.82 (82.38)	Nearly perfect
Absolute H:Q ratio	0.024 (-0.307-0.350)	0.889	0.00 (0.07)	Trivial	0.553 (0.274-0.746)	<0.001	0.31 (30.58)	Large
Relative H:Q ratio	0.298 (-0.034-0.571)	0.007	0.08 (8.88)	Small	0.796 (0.632-0.891)	<0.001	0.63 (63.44)	Very Large
Angle of Crossover	-0.008 (-0.336-0.321)	0.961	0.00 (0.00)	Trivial	0.457 (0.151-0.683)	0.005	0.21 (21.13)	Moderate

In contrast, relative BF_{LH} FL demonstrated moderate correlations with absolute hamstring eccentric peak torque and the angle of crossover. Additionally, large and very large correlations were observed between relative BF_{LH} FL and absolute H:Q ratio and relative H:Q ratio, respectively (Figure 5-4 A & B). A nearly perfect relationship was observed between relative BF_{LH} FL and relative hamstring eccentric peak torque (Figure 5-4 C).

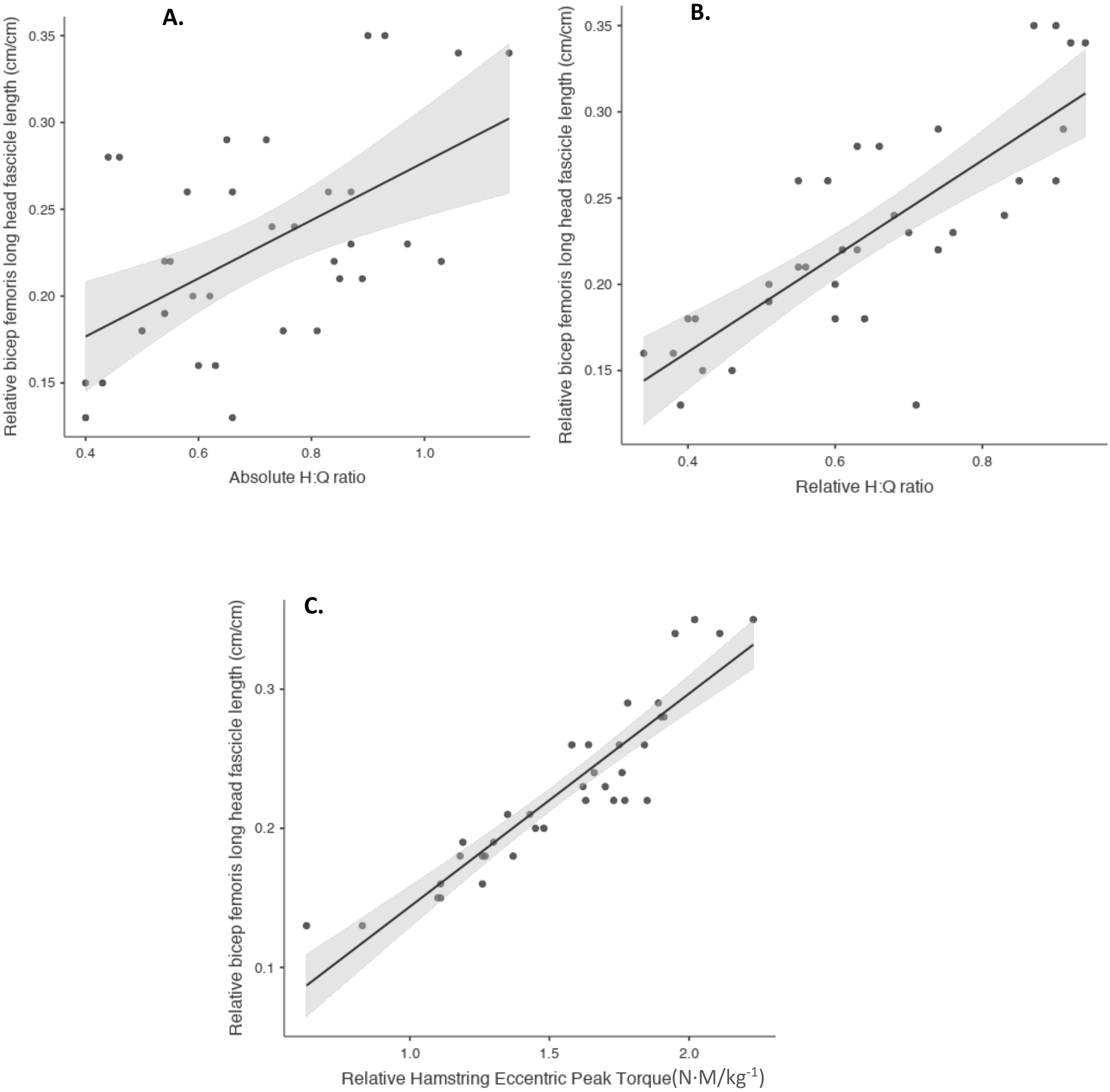


Figure 5-4 A. Relationship between relative bicep femoris fascicle length and absolute H:Q ratio, B. Relationship between relative bicep femoris fascicle length and Relative H:Q ratio and C. Relationship between relative bicep femoris fascicle length and relative.

5.1.3.4 High and low risk groupings

Independent t-tests of muscle architecture and isokinetic hamstring eccentric peak torques between groups of perceived high- and low-risk of future HSI demonstrated significant and large differences for both absolute and relative parameters ($p < 0.05$, $d > 1.5$) (Table 5-5). As there was a nearly perfect relationship observed between the relative hamstring eccentric peak torque and relative BF_{LH} FL (Figure 5-5), a qualitative score was applied to each relative measurement to define the two groups (i.e., high and low risk) from a combination of both hamstring eccentric peak torque and relative BF_{LH} FL.

Table 5-5 Between perceived high- and low-risk groups for muscle architecture characteristics and isokinetic hamstring eccentric peak torque.

Perceived risk of future HSI		**High**	**Low**	***p***	**ES (*d*) *(95% CI)***	**Effect size Descriptor**
Modifiable Risk factor parameter	**Limb**	**Mean (SD)**				
Absolute BF_{LH} Fascicle Length (cm)	Left	8.61 (0.72)	11.38 (1.52)	<0.001	2.32 (1.05-3.59)	Large
	Right	8.75 (0.93)	11.50 (1.53)	<0.001	2.17 (0.94-3.41)	Large
Relative BF_{LH} Fascicle Length	Left	0.17 (0.02)	0.28 (0.05)	<0.001	2.95 (1.53-4.36)	Large
	Right	0.17 (0.03)	0.28 (0.05)	<0.001	2.87 (1.48-4.27)	Large
Absolute hamstring eccentric peak torque (N·m)	Left	107.25 (22.80)	166.88 (12.77)	<0.001	3.23 (1.40-3.93)	Large
	Right	115.83 (18.70)	176.96 (15.87)	<0.001	3.52 (1.59-4.24)	Large
Relative hamstring eccentric peak torque (N·m/kg^{-1})	Left	1.16 (0.25)	1.81 (0.14)	<0.001	3.23 (1.38-3.92)	Large
	Right	1.25 (0.20)	1.92 (0.17)	<0.001	3.52 (1.54-4.17)	Large

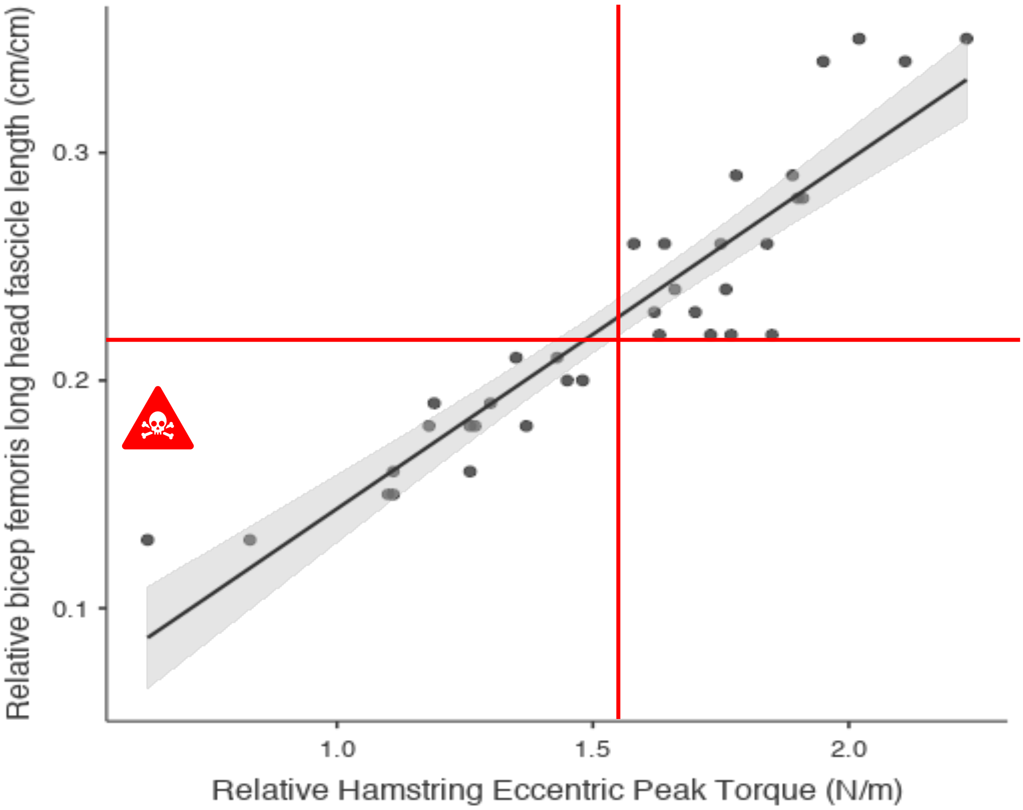


Figure 5-5 Individual relative peak torque and relative bicep femoris fascicle length values with proposed thresholds to group between long and strong – short and weak. In the style of "Quadrant of doom" (Timmins et al. 2016).

5.1.3.5 Kinematic and EMG reliability running data

5.1.3.6 Between-stride reliability

Acceptable levels of absolute reliability (CV <10%) was observed for all peak kinematic and EMG variables between three observed strides for both the left and right limbs (Table 5-6 & 5-8). High to nearly perfect (lower bound 95% CI, 0.50-0.99) ICCs were observed for all kinematic and EMG variables (Table 5-7 & 5-9). Both the absolute and relative between-stride reliability demonstrated a general trend of decreasing variability as running velocity increased.

Between group stride variability (ICCs) across three consecutive strides was also determined (appendix two), highlighting that the high-risk group also had greater variability across all running speeds for hip and knee extension-flexion and change in knee angular velocity. With a similar degree of variability for estimated BF_{LH} MTU length and take-off.

Table 5-6 Between stride absolute reliability (CV [95%CI]) for kinematic variables for the left and right limbs											
LEFT						**RIGHT**					
	Velocity (km·hr¹)						**Velocity (km·hr¹)**				
	8	**10**	**12**	**14**	**16**		**8**	**10**	**12**	**14**	**16**
Peak Knee flexion	9.11 (9.01-9.20)	7.09 (6.95-7.23)	3.45 (3.37-3.53)	3.73 (3.61-3.85)	3.12 (2.95-3.29)	**Peak Knee flexion**	9.65 (9.56-9.74)	6.02 (5.88-6.16)	5.34 (5.26-5.42)	4.64 (4.52-4.76)	2.74 (2.57-2.91)
Peak knee extension	9.04 (8.95-9.13)	9.67 (9.53-9.81)	8.46 (8.38-8.54)	8.74 (8.62-8.86)	8.39 (8.22-8.56)	**Peak knee extension**	9.53 (9.44-9.62)	9.78 (9.64-9.92)	8.39 (8.31-8.47)	8.65 (8.53-8.77)	7.86 (7.69-8.03)
Peak hip flexion	7.01 (6.92-7.10)	6.30 (6.16-6.44)	5.53 (5.45-5.61)	3.30 (3.18-3.42)	4.32 (4.15-4.49)	**Peak hip flexion**	9.57 (9.48-9.66)	9.54 (9.40-9.68)	9.66 (9.58-9.74)	6.29 (6.17-6.41)	7.08 (6.91-7.25)
Peak hip extension	7.33 (7.23-7.43)	6.53 (6.38-6.66)	5.21 (5.10-5.30)	3.84 (3.72-3.96)	3.67 (3.50-3.84)	**Peak hip extension**	6.72 (6.63-6.81)	8.57 (8.43-8.71)	5.95 (5.87-6.03)	5.65 (5.53-5.77)	3.89 (3.72-4.06)
Change in knee angular velocity	12.88 (10.33 – 15.22)	9.53 (8.28 – 10.89)	8.56 (7.38 – 9.81)	7.21 (6.04 – 8.35)	5.69 (3.88 – 7.49)	**Change in knee angular velocity**	10.22 (8.80 – 11.60)	9.68 (8.46 – 10.97)	8.75 (7.52 – 10.01)	7.33 (6.12 – 8.55)	5.01 (3.46 – 6.40)
Peak BF Muscle tendon unit length	0.92 (0.83-1.01)	0.92 (0.78-1.06)	1.81 (1.73-1.89)	0.91 (0.79-1.03)	1.80 (1.63-1.97)	**Peak BF Muscle tendon unit length**	0.91 (0.82-1.00)	0.92 (0.78-1.06)	0.91 (0.83-0.99)	0.90 (0.78-1.02)	0.90 (0.73-1.07)
Take off	3.96 (3.41-4.50)	4.62 (4.20-5.03)	3.39 (3.08-3.70)	4.31 (3.79-4.83)	3.91 (3.46-4.37)	**Take off**	4.06 (3.81-4.32)	4.76 (4.27-5.25)	3.13 (2.73-3.52)	3.94 (3.51-4.36)	4.46 (3.93-4.99)

Table 5-7 Between stride relative reliability (ICC [95% CI]) for kinematic variables for the left and right limbs

LEFT						**RIGHT**					
	Velocity (km·hr¹)						**Velocity (km·hr¹)**				
	8	**10**	**12**	**14**	**16**		**8**	**10**	**12**	**14**	**16**
Peak Knee flexion	0.706 (0.534-0.836)	0.870 (0.728 - 0.923)	0.865 (0.705-0.956)	0.885 (0.811-0.932)	0.914 (0.853-0.967)	**Peak Knee flexion**	0.696 (0.554-0.877)	0.716 (0.686 - 0.775)	0.785 (0.724-0.806)	0.806 (0.745-0.872)	0.869 (0.803-0.901)
Peak knee extension	0.724 (0.569 - 0.884)	0.689 (0.562-0.797)	0.730 (0.574-0.865)	0.789 (0.602-0.888)	0.809 (0.672-0.901)	**Peak knee extension**	0.643 (0.589 - 0.884)	0.690 (0.627-0.757)	0.733 (0.674-0.775)	0.791 (0.702-0.868)	0.819 (0.726-0.871)
Peak hip flexion	0.779 (0.624 - 0.864)	0.803 (0.697-0.904)	0.834 (0.710-0.922)	0.868 (0.798-0.909)	0.921 (0.875-0.959)	**Peak hip flexion**	0.793 (0.724 - 0.864)	0.812 (0.767-0.864)	0.828 (0.780-0.852)	0.867 (0.803-0.899)	0.869 (0.810-0.900)
Peak hip extension	0.845 (0.717-0.903)	0.820 (0.703-0.897)	0.881 (0.754-0.935)	0.913 (0.848-0.967)	0.902 (0.821-0.954)	**Peak hip extension**	0.815 (0.757-0.873)	0.826 (0.763-0.884)	0.867 (0.798-0.905)	0.901 (0.878-0.927)	0.902 (0.877-0.954)
Change in knee angular velocity	0.805 (0.702-0.893)	0.812 (0.699-0.877)	0.861 (0.744-0.955)	0.893 (0.812-0.954)	0.872 (0.801-0.914)	**Change in knee angular velocity**	0.795 (0.723-0.842)	0.802 (0.738-0.874)	0.866 (0.799-0.909)	0.890 (0.816-0.957)	0.912 (0.870-0.960)
Peak BF Muscle tendon unit length	0.912 (0.859-0.972)	0.920 (0.886-0.962)	0.945 (0.911-0.977)	0.983 (0.965-0.993)	0.953 (0.921-0.982)	**Peak BF Muscle tendon unit length**	0.945 (0.904-0.973)	0.940 (0.906-0.942)	0.939 (0.910-0.967)	0.933 (0.905-0.972)	0.962 (0.931-0.992)
Take off	0.915 (0.886-0.954)	0.889 (0.869-0.924)	0.902 (0.824-0.948)	0.910 (0.885-0.940)	0.909 (0.891-0.934)	**Take off**	0.914 (0.878-0.943)	0.910 (0.893-0.944)	0.922 (0.884-0.946)	0.930 (0.905-0.950)	0.919 (0.879-0.946)

Table 5-8 Between stride absolute reliability (CV [95%CI]) for EMG variables for the left and right limbs

LEFT						RIGHT					
	Velocity (km·hr¹)						**Velocity (km·hr¹)**				
	8	**10**	**12**	**14**	**16**		**8**	**10**	**12**	**14**	**16**
Peak BF	6.11 (5.41-6.60)	6.09 (5.75-6.44)	6.39 (5.99-6.80)	6.40 (6.01-6.79)	5.15 (4.75-5.52)	**Peak BF**	6.13 (5.58-6.72)	6.01 (5.68-6.51)	6.10 (5.88-6.32)	6.29 (5.81-6.99)	5.88 (5.52-6.02)
Peak ST	8.14 (7.75-8.63)	7.67 (7.23-8.01)	8.06 (7.78-8.34)	7.74 (7.42-8.06)	6.39 (5.99-6.77)	**Peak ST**	8.53 (8.24-8.82)	9.80 (9.24-10.34)	8.19 (7.81-8.61)	9.65 (9.23-10.07)	6.80 (6.29-7.33)
BF:ST	7.91 (7.12-8.72)	7.94 (7.46-8.41)	7.62 (7.05-8.18)	7.02 (6.58-6.42)	7.32 (7.01-7.69)	**BF:ST**	7.57 (7.08-8.06)	7.54 (7.12-7.98)	7.66 (7.08-8.04)	7.29 (6.87-7.71)	7.16 (6.81-7.45)

Table 5-9 Between stride relative reliability (ICC [95% CI]) for EMG variables for the left and right limbs

LEFT						RIGHT					
	Velocity (km·hr¹)						**Velocity (km·hr¹)**				
	8	**10**	**12**	**14**	**16**		**8**	**10**	**12**	**14**	**16**
Peak BF	0.777 (0.714-0.833)	0.830 (0.788 -0.883)	0.765 (0.705-0.826)	0.785 (0.771-0.802)	0.814 (0.773-0.857)	**Peak BF**	0.706 (0.664-0.767)	0.786 (0.746 - 0.825)	0.755 (0.704-0.805)	0.746 (0.705-0.782)	0.809 (0.773-0.855)
Peak ST	0.694 (0.590 - 0.784)	0.709 (0.662-0.767)	0.710 (0.674-0.755)	0.709 (0.672-0.748)	0.793 (0.732-0.851)	**Peak ST**	0.713 (0.679 - 0.754)	0.694 (0.647 - 0.747)	0.703 (0.644 - 0.765)	0.782 (0.722-0.843)	0.789 (0.736-0.841)
BF:ST	0.719 (0.684 - 0.745)	0.801 (0.757- 0.844)	0.734 (0.708-0.772)	0.706 (0.798-0.909)	0.821 (0.775-0.860)	**BF:ST**	0.713 (0.674 - 0.762)	0.712 (0.877-0.754)	0.728 (0.680-0.762)	0.767 (0.703-0.809)	0.819 (0.772-0.861)

5.1.3.7 Between group measurements – running velocity, kinematic and electromyography differences

Between high- and low-risk groups, no significant differences were identified for the maximal running velocities achieved within normalization between high and low risk groups, with only trivial effect sizes observed in favour of the low-risk group (Table 5-10).

Table 5-10 Between perceived high- and low-risk groups for maximal running velocity achieved during the EMG normalization

Perceived risk of future HSI	High	Low	*p*	ES (*d*) (95% CI)	Descriptor
Modifiable Risk factor parameter	Mean (SD) ($km{\cdot}hr^{-1}$)				
Combination of relative BF_{LH} Fascicle length and Relative eccentric peak torque	24.75 (1.96)	25.13 (1.92)	0.705	0.19 (-1.71-2.45)	Trivial

For peak kinematic variables (Table 5-11 & 5-12), significant and meaningful differences between high- and low-risk groups (which were consistent between limbs), were observed for peak hip extension, relative take off. Further, non-significant but small and moderate consistent differences were observed for peak change in knee angular velocity. Peak EMG variables (Table 5-13) demonstrated significant and meaningful differences for relative BF activation across the greater running speeds (12-, 14- and 16 $km{\cdot}hr^{-1}$), with small non-significant differences in relative ST activation across the same running velocities. This resulted in large and significant differences in BF:ST ratio at the greater running velocities (14- and 16 $km{\cdot}hr^{-1}$). All differences within EMG were greater within the high-risk group when compared to the low-risk group.

Table 5-11 Peak kinematic differences observed for the left limb across all running velocities.

Variable	RISK	8 km·hr⁻¹ Mean (SD)	8 km·hr⁻¹ p	8 km·hr⁻¹ d (95% CI)	10 km·hr⁻¹ Mean (SD)	10 km·hr⁻¹ p	10 km·hr⁻¹ d (95% CI)	12 km·hr⁻¹ Mean (SD)	12 km·hr⁻¹ p	12 km·hr⁻¹ d (95% CI)	14 km·hr⁻¹ Mean (SD)	14 km·hr⁻¹ p	14 km·hr⁻¹ d (95% CI)	16 km·hr⁻¹ Mean (SD)	16 km·hr⁻¹ p	16 km·hr⁻¹ d (95% CI)
Hip Extension (°)	High	-5.78 (3.15)	0.014	1.14 (-0.14 - 2.13)	-8.22 (6.18)	0.002	1.66 (0.51 - 2.78)	-10.57 (5.24)	0.010	1.63 (0.50 - 2.76)	-12.33 (7.04)	0.010	1.84 (0.67 - 3.01)	-13.48 (3.38)	0.009	1.87 (0.69 - 3.04)
	Low	1.90 (3.66)			0.04 (3.33)			-2.55 (4.59)			-4.73 (4.23)			-6.01 (4.54)		
Hip Flexion (°)	High	32.17 (5.49)	0.013	1.24 (0.17 - 2.31)	39.91 (4.84)	0.069	0.79 (-0.23 - 1.80)	45.07 (4.34)	0.034	0.99 (-0.05 - 2.03)	49.66 (5.16)	0.464	0.10 (-0.88 - 1.08)	53.31 (5.75)	0.355	0.15 (-0.83 - 1.13)
	Low	39.20 (5.88)			44.35 (6.34)			49.54 (4.69)			50.20 (5.70)			54.24 (6.57)		
Knee Extension (°)	High	5.14 (4.57)	0.467	0.27 (-0.66 - 1.19)	5.07 (4.78)	0.464	0.27 (-0.66 - 1.20)	4.70 (3.61)	0.706	0.24 (-0.69 - 1.16)	3.71 (3.55)	0.360	0.51 (-0.44 - 1.44)	3.66 (3.05)	0.756	0.47 (-0.47 - 1.41)
	Low	6.33 (4.28)			6.27 (3.91)			5.56 (3.50)			5.46 (3.22)			5.15 (3.23)		
Knee Flexion (°)	High	80.50 (11.92)	0.237	0.37 (-0.62 - 1.36)	89.35 (2.94)	0.490	0.46 (-0.48 - 1.39)	99.39 (3.69)	0.454	0.49 (-0.45 - 1.42)	105.13 (4.18)	0.643	0.54 (-0.42 - 1.47)	110.34 (5.65)	0.652	0.49 (-0.50 - 1.49)
	Low	76.29 (10.93)			87.53 (4.70)			97.71 (3.09)			102.42 (5.81)			107.79 (4.62)		
Change in Knee Angular Velocity (deg/s)	High	3.45 (0.37)	0.229	0.62 (-0.33 - 1.56)	4.01 (0.41)	0.201	0.67 (-0.28 - 1.62)	4.16 (0.35)	0.127	0.79 (-0.17 - 1.75)	4.35 (0.45)	0.097	0.90 (-0.07 - 1.87)	4.64 (0.52)	0.156	0.76 (0.20 - 1.71)
	Low	3.23 (0.34)			3.66 (0.61)			3.89 (0.33)			3.98 (0.37)			4.34 (0.21)		
Muscle tendon unit	High	1.09 (0.02)	0.055	0.63 (-0.37 - 1.64)	1.10 (0.02)	0.112	0.63 (-0.37 - 1.64)	1.11 (0.02)	0.121	0.00 (-0.98 - 0.98)	1.11 (0.02)	0.542	0.00 (-0.98 - 0.98)	1.12 (0.02)	0.134	0.00 (-0.98 - 0.98)
	Low	1.10 (0.01)			1.11 (0.01)			1.11 (0.01)			1.11 (0.03)			1.12 (0.01)		
Take off (% gait)	High	11.52 (4.81)	0.173	0.72 (-0.29 - 1.73)	10.92 (3.54)	0.021	1.31 (0.23 - 2.39)	9.02 (2.63)	0.001	2.06 (0.85 - 3.27)	9.67 (3.80)	0.015	1.38 (0.29 - 2.47)	10.09 (1.98)	0.012	1.61 (0.48 - 2.73)
	Low	14.94 (4.72)			14.74 (2.14)			13.76 (1.92)			14.11 (2.49)			14.11 (2.93)		

Significant differences (<0.05) are denoted by [shaded cell]. The magnitude of the difference (*d*) is denoted by: Trivial (< 0.35) | Small (0.35 - 0.80) | Moderate (0.80 - 1.50) | Large (>1.51)

Table 5-12 Peak kinematic differences observed for the right limb across all running velocities.

Variable	RISK	8 km·hr^{-1}			10 km·hr^{-1}			12 km·hr^{-1}			14 km·hr^{-1}			16 km·hr^{-1}		
		Mean (SD)	*p*	*d* (95% CI)	Mean (SD)	*p*	*d* (95% CI)	Mean (SD)	*p*	*d* (95% CI)	Mean (SD)	*p*	*d* (95% CI)	Mean (SD)	*p*	*d* (95% CI)
Hip Extension (°)	High	-5.45 (3.32)	0.009	0.67 (-0.34 - 1.67)	-7.86 (6.24)	0.002	1.30 (0.22 - 2.38)	-9.48 (5.44)	0.012	1.52 (0.40 - 2.63)	-10.86 (6.05)	0.006	1.45 (0.35 - 2.55)	-12.13 (3.17)	0.005	1.94 (0.75 - 3.13)
	Low	2.73 (4.73)			1.10 (3.90)			-1.39 (5.23)			-2.98 (4.77)			-3.78 (5.19)		
Hip Flexion (°)	High	32.47 (5.49)	0.073	0.79 (-0.22 - 1.81)	39.85 (5.93)	0.306	0.26 (-0.72 - 1.24)	44.08 (6.56)	0.276	0.38 (-0.61 - 1.37)	49.32 (5.68)	0.196	0.57 (-0.43 - 1.57)	54.33 (6.34)	0.477	0.03 (-0.95 - 1.01)
	Low	37.50 (7.09)			41.81 (8.88)			46.59 (6.63)			52.32 (4.73)			54.56 (9.32)		
Knee Extension (°)	High	4.60 (4.79)	0.311	0.47 (-0.48 - 1.40)	4.17 (4.18)	0.137	0.57 (-0.43 - 1.57)	4.48 (3.86)	0.206	0.42 (-0.57 - 1.41)	3.84 (3.50)	0.563	0.45 (-0.49 - 1.38)	3.33 (3.65)	0.252	0.59 (-0.36 - 1.53)
	Low	6.73 (4.12)			6.82 (5.09)			6.12 (3.88)			5.50 (3.88)			6.10 (5.59)		
Knee Flexion (°)	High	80.48 (8.52)	0.335	0.51 (-0.44 - 1.44)	89.75 (8.44)	0.627	0.49 (-0.46 - 1.42)	98.22 (5.10)	0.581	0.51 (-0.42 - 1.44)	106.14 (4.13)	0.575	0.55 (-0.40 - 1.48)	113.12 (5.38)	0.778	0.95 (-0.09 - 1.98)
	Low	76.60 (6.62)			85.61 (8.28)			95.22 (5.37)			104.06 (3.33)			107.89 (5.65)		
Change in Knee Angular Velocity (deg/s)	High	3.50 (0.30)	0.067	0.99 (0.00 - 1.96)	4.20 (0.65)	0.066	1.00 (0.01 - 1.97)	4.31 (0.43)	0.182	0.70 (-0.25 - 1.65)	4.59 (0.61)	0.129	0.81 (-0.15 - 1.77)	4.56 (0.45)	0.053	1.06 (0.07 - 2.04)
	Low	3.23 (0.25)			3.69 (0.33)			4.01 (0.43)			4.16 (0.44)			4.10 (0.42)		
Muscle tendon unit	High	1.09 (0.02)	0.164	0.63 (-0.37 - 1.64)	1.10 (0.02)	0.458	0.00 (-0.98 - 0.98)	1.11 (0.02)	0.393	0.00 (-0.98 - 0.98)	1.11 (0.02)	0.213	0.63 (-0.31 - 1.58)	1.12 (0.02)	0.485	0.00 (-0.98 - 0.98)
	Low	1.10 (0.01)			1.10 (0.02)			1.11 (0.01)			1.12 (0.01)			1.12 (0.02)		
Take off (% gait)	High	12.06 (3.89)	0.166	0.71 (-0.29 - 1.73)	11.16 (3.59)	0.018	1.34 (0.26 - 2.43)	10.66 (3.24)	0.009	1.30 (0.22 - 2.37)	10.33 (2.99)	0.034	1.18 (0.12 - 2.24)	10.27 (2.80)	0.009	1.50 (0.39 - 2.61)
	Low	14.99 (4.28)			15.40 (2.66)			13.19 (3.06)			13.94 (3.13)			15.43 (3.96)		

Significant differences (<0.05) are denoted by | The magnitude of the difference (*d*) is denoted by: Trivial (< 0.35) | Small (0.35 - 0.80) | Moderate (0.80 - 1.50) | Large (>1.51)

Table 5-13 Peak EMG differences for the left and right legs across all running velocities.

Limb	Variable	Risk	8 $km{\cdot}hr^{-1}$			10 $km{\cdot}hr^{-1}$			12 $km{\cdot}hr^{-1}$			14 $km{\cdot}hr^{-1}$			16 $km{\cdot}hr^{-1}$		
			Mean (SD)	*p*	*d (95%CI)*	Mean (SD)	*p*	*d (95%CI)*	Mean (SD)	*p*	*d (95%CI)*	Mean (SD)	*p*	*d (95%CI)*	Mean (SD)	*p*	*d (95%CI)*
Left	Relative BF (%)	HIGH	47.74 (11.51)	0.165	0.49 (-0.44 - 1.43)	58.89 (13.74)	0.352	0.17 (-0.44 - 1.43)	63.34 (12.70)	0.208	0.42 (-0.44 - 1.43)	67.86 (15.46)	0.008	1.17 (0.14 – 2.03)	74.40 (9.29)	0.008	1.17 (0.22 - 1.73)
		LOW	41.91 (12.11)			51.59 (13.57)			56.42 (3.60)			59.42 (12.96)			62.92 (8.37)		
	Relative ST (%)	HIGH	51.21 (7.09)	0.522	0.13 (-0.67 - 0.48)	61.69 (13.65)	0.384	0.07 (0.37 - 0.58	69.16 (10.19)	0.202	0.33 (0.97 - 3.28	72.83 (8.94)	0.132	0.56 (-0.22 - 1.28)	78.45 (10.00)	0.195	0.66 (-0.20 - 1.20)
		LOW	52.58 (10.16)			60.76 (12.20)			66.15 (12.94)			67.23 (10.50)			72.12 (13.60)		
	BF:ST activation	HIGH	0.93 (0.80)	0.361	0.15 (-0.54 - 0.55)	0.95 (0.68)	0.384	0.08 (-0.54 - 0.73)	0.92 (0.63)	0.392	0.01 (-0.54 - 0.55)	0.93 (0.14)	0.001	1.53 (0.74 - 2.33)	0.95 (0.59)	0.000	3.78 (3.24 - 4.33)
		LOW	0.80 (0.49)			0.85 (0.64)			0.85 (0.59)			0.88 (0.61)			0.87 (0.29)		
Right	Relative BF (%)	HIGH	40.8 (13.49)	0.375	0.11 (-0.38 - 0.55)	56.11 (10.33)	0.144	0.54 (-0.18 – 1.21)	65.12 (13.47)	0.215	0.41 (-0.38 - 0.82)	67.27 (8.95)	0.019	1.00 (0.22 - 1.76)	77.92 (12.85)	0.008	1.16 (0.16 - 2.16)
		LOW	39.38 (13.31)			50.33 (11.24)			56.95 (11.55)			59.44 (6.60)			70.07 (10.92)		
	Relative ST (%)	HIGH	48.87 (12.74)	0.392	0.00 (0.99 - 1.00)	60.23 (15.46)	0.372	0.12 (-0.81 - 1.04)	69.74 (12.96)	0.250	0.35 (-0.58 - 1.28)	70.45 (13.52)	0.166	0.49 (-0.44 - 1.43)	79.18 (9.87)	0.192	0.45 (-0.31 - 1.58)
		LOW	48.88 (9.97)			61.97 (13.89)			64.59 (9.97)			66.14 (7.02)			75.64 (10.92)		
	BF:ST activation	HIGH	0.83 (0.51)	0.374	0.11 (-0.38 - 0.55)	0.93 (0.43)	0.391	0.02 (-0.47 - 0.49)	0.93 (0.52)	0.392	0.01 (-0.64 - 0.65)	0.95 (0.07)	0.000	3.07 (1.50 - 4.37)	0.98 (0.40)	0.000	2.93 (2.09 - 3.61)
		LOW	0.81 (0.62)			0.81 (0.47)			0.88 (0.63)			0.90 (0.06)			0.93 (0.16)		
Significant difference (<0.05) are denoted by					The magnitude of differences (*d*) are denoted by:				Trivial (<0.35)		Small (0.35 - 0.80)			Moderate (0.80 - 1.50)		Large (>1.51)	

5.1.3.7.1 Post-hoc power analysis

Post-hoc power analysis was performed using Jamovi Jpower tool (Jamovi project (2018) Computer Software, retrieved from https://www.jamovi.org) to determine the statistical power achieved within the present study; it was highlighted that effect sizes between 1.51 – 1.94 had an 80-95% power to detect true effects. Whereas effect sizes of greater than 1.94 had a 95% power to detect change. This indicates that it is highly likely that the trivial-moderate differences observed were underpowered, with good-likely chances of missing.

5.1.4 Discussion

5.1.4.1 Modifiable risk factors

Similar to previous research US and isokinetic measures are highly reliable, with acceptable levels of variability (CV < 10%) and very large to nearly perfect levels of relative reliability (ICC >0.8) (Graham-Smith, et al., 2013; Timmins, et al., 2015). Highlighting that both methods can be used reliably to screen an individual's HSI risk factors, eccentric strength and muscle architecture. To date, no study has looked to observe if associations exist between the modifiable risk factors (eccentric hamstring strength and BF_{LH} FL). The data in the present study only partially met the hypothesis, as absolute BF_{LH} FL demonstrated small and trivial associations with all isokinetic measures (Table 5-4). In contrast, however, relative BF_{LH} FL demonstrated moderate to nearly perfect associations with isokinetic measures (Table 5-4). Relative BF_{LH} FL was able to explain up 82.38% of the variance of relative eccentric isokinetic hamstring strength. This finding more than likely explains the very large association of relative BF_{LH} FL observed with other variables such as, relative H:Q ratio.

To the researcher's knowledge, this is the first-time associations have been observed between eccentric hamstring strength and BF_{LH} FL. Previous literature has identified that there is a decrease in HSI risk, with increases in both eccentric hamstring strength and BF_{LH} FL (Bourne, et al., 2015; Opar, et al., 2015; Timmins, Bourne, et al., 2016; Timmins, et al., 2015). This has been visually presented using the aptly named "quadrant of doom"– whereby subjects presenting low levels of eccentric strength and BF_{LH} FL were at a greater likelihood of HSI occurrence (Bourne, et al., 2018). However, the "quadrant of doom" could be misleading with several caveats (including the one presented within the present thesis), as there were individuals who displayed a high level of eccentric strength and low BF_{LH} FL. Secondly, previously presented normative data of reduced risk of future HSI for both eccentric hamstring strength and BF_{LH} FL, have all used absolute values, this is despite bodyweight able to explain up to ¼ of eccentric hamstring strength when using the Nordbord device (Bourne, et al., 2018; Buchheit, et al., 2017). Furthermore, the data presented within the present study indicates that relative measures are extremely closely associated and could be more appropriate. However, as this is the first study to observe such associations, there is minimal normative data established which could indicate an increase in HSI risk.

Within the present study, participants were able to be differentiated as high and low risk groups – using a combination of relative eccentric hamstring strength and relative BF_{LH} FL – due to the nearly perfect associations between these measures. Additionally, the grouping of high and low risk was identical between limbs, where large and significant differences were identified between groups (Table 5-5), which allowed for all strength and FL to be used to assess any possible differences within EMG and kinematics between groups. Consistent with previous research, the high-risk group regarding absolute BF_{LH} FL were lower for both the left and right limbs than a previously established normative value (10.56 cm) for athletic populations (Bourne, et al., 2018; Timmins, Bourne, et al., 2016), with the high-risk group up to 4.1 times more likely to sustain a future HSI. Although, no values for an elevated risk of HSI incidence have been established for eccentric strength using isokinetic strength assessment, with only limited predictive ability for future HSIs (Green, et al., 2018). When compared to normative data, the absolute eccentric hamstring strength for the low risk group (within the present study), were greater than that of elite sprinters and under-20 soccer midfielders and forwards (Guex, et al., 2012). However, both groups within the present study were considerably lower when compared to under-20 soccer defenders (Costa, Detanico, Dal Pupo, & la Rocha Freitas, 2015). Furthermore, the mean peak eccentric torque for the entire sample within the present study were lower than data presented for both uninjured team sport athletes and those with a previous HSI (Australian footballers and soccer players) (Bennell, et al., 1998; van Dyk, Bahr, et al., 2018). Therefore, it should be highlighted that the eccentric isokinetic measurements made within the present study may not indicate high or low risk of sustaining a future HSI, but it has allowed for characterization of participants within the present study with regards to the hamstring's force generating capacity, which is a modifiable risk factor.

5.1.4.2 Kinematic and Electromyography

5.1.4.3 Reliability

The data presented within the current study, demonstrates acceptable levels of absolute and relative reliability between three consecutive strides for all kinematic and EMG variables across all running velocities (Table 5-6 – 5-9). Additionally, there was an increase in reliability as running velocity increased, indicating that the consistency of running strategy and performance improved across the entire sample with greater running velocities. Interestingly, the high-risk group had greater running variability across all running velocities for kinematic measures (Appendix two), potentially further supporting the high-risk observations. With increased running variability, especially in the pelvis region (anterior-posterior pelvic tilt), there is the potential increased risk of HSI occurrence (Schuermans, Tiggelen, et al., 2017). Across the sample, the kinematic data demonstrated similar trends for both limbs - with significant increases in peak knee flexion, peak hip flexion angle, peak estimated MTU length and EMG measures, with increased running velocity.

Consistent with previous running research, intrasession reliability for kinematic measures within the present study were very high with acceptable levels of variability (Girard, Brocherie, Morin, & Millet, 2016), even though the running velocities within the present study were considerably lower; maximum of 4.4 $m \cdot s^{-1}$ within the present study compared to >6.30 $m \cdot s^{-1}$ (Girard, et al., 2016). This finding demonstrates that there was a minimal impact of familiarization on the kinematic variables explored, similar to the results observed by Meyer et al. (2019), whereby hip and knee ROM did not require an acclimatization procedure for treadmill walking gait. Although Meyer and colleagues (2019) identified that stride characteristics (e.g. stride length, stride time etc.) did require an acclimatization window of over 100 strides, however the authors utilised an alternative method of identifying TD and TO within the gait cycle compared to the present study (Meyer, et al., 2019). Although the submaximal running velocities used within the present study did not reach maximum speeds associated with HSI, it would be expected that the trend of a reduction in movement variability may continue as velocity increases beyond those studied here. However, it has been highlighted that a possible cause of HSI is a lack of running coordination, specifically around the trunk (pelvis and core) (Schuermans, Tiggelen, et al., 2017), which could be due to a systematic increase in movement variability when reaching maximum running velocities (Schuermans, Tiggelen, et al., 2017).

5.1.4.4 Between group peak kinematic and EMG differences

In agreement with the hypotheses, there were significant and meaningful differences observed between high and low risk groups across all running velocities for both peak kinematic and EMG variables (Table 5-11 – 5-13). Non-significant and trivial differences were observed between groups for maximal running velocity when normalizing EMG. Indicating that the differences highlighted between high and low risk groups were not influenced by the maximal running ability of the participants. Therefore, it would be presumed for kinematic and EMG differences to be related to the measured risk factors; a meaningful difference would have to be consistent between limbs, instead of individual variability. Therefore, these differences and trends will form the focus of this discussion.

5.1.4.5 Kinematic differences

Significant and meaningful differences were identified in peak hip extension (greater for high-risk group) and relative take-off time (greater for low-risk group), which were consistent between limbs. Furthermore, non-significant, small and moderate differences were also observed for knee joint kinematics including, peak knee flexion, extension and change in knee angular velocity, which was consistent between limbs. Differences in peak hip flexion and MTU length were trivial or small, non-significant with no consistency between limbs. As a visual representation of the kinematic differences an angle-angle plot was produced (Figure), where mean hip- and knee-angles across the gait cycle were plotted simultaneously, for both high and low-risk groups.

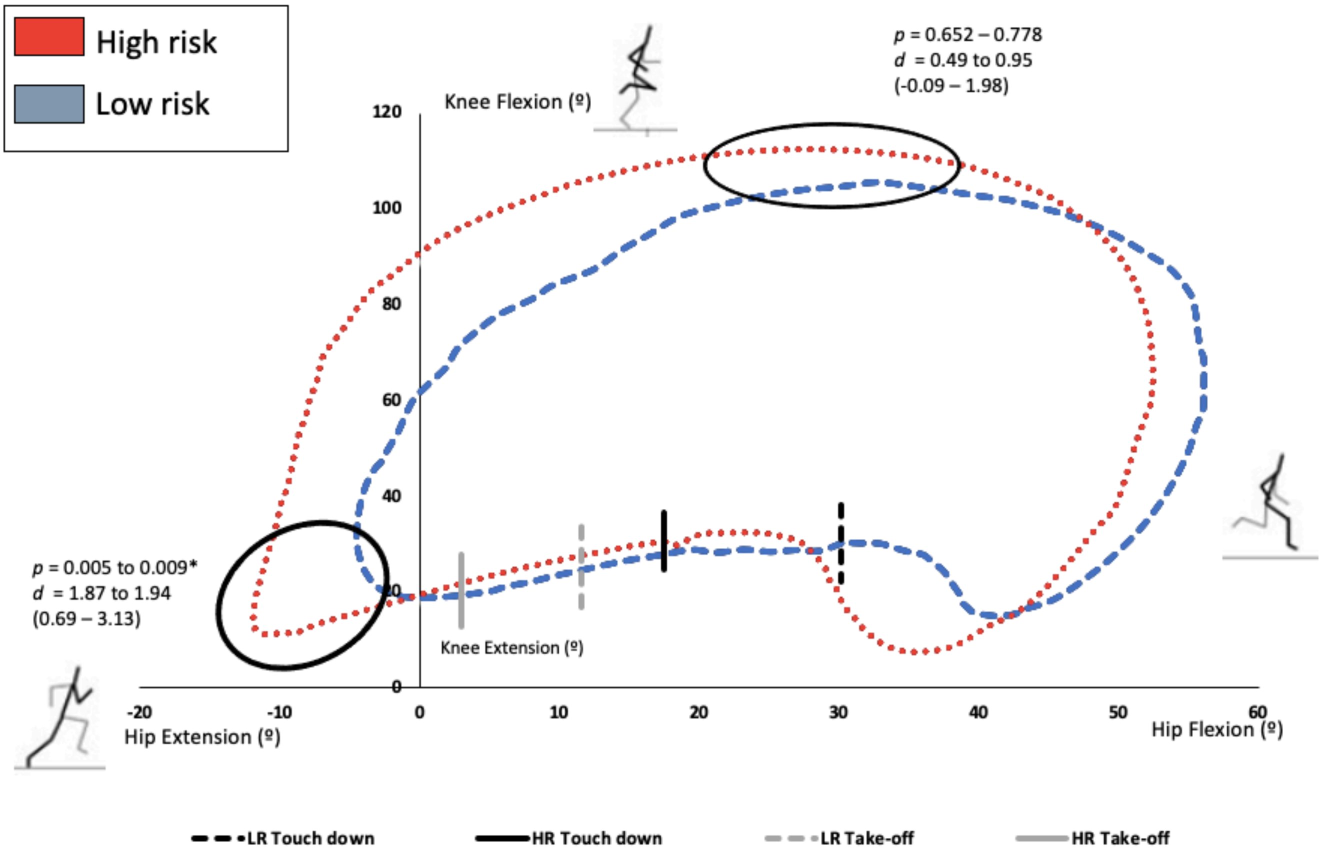


Figure 5-6 Hip and Knee angle-angle plot highlighting the mean kinematic differences between high and low risk groups across the entire gait cycle.

The present study demonstrates novel information regarding the effect of eccentric hamstring strength and relative BF_{LH} FL on running gait. Specifically, participants who were identified as high risk (weaker and shorter FL), were found to potentially be compensating their temporal gait characteristics. The observed results demonstrate that the high-risk group move through a greater hip extension, while the low-risk group consistently TO later as a percent of their gait cycle. A possible explanation could lie in the fact that a muscle's working range is reduced due to a decreased BF_{LH} FL, therefore higher risk individuals could be compensating by having to apply force across a larger ROM more rapidly. The consistent differences within the running gait with hip extension, could indicate possible limits with pelvic coordination, specifically the potential for anterior pelvic tilt to increase the larger range of hip extension. Similarly, Small, et al. (2009) identified that under fatigued conditions, when the risk of HSI is greater and it would be expected there would be reduced strength, soccer players demonstrated a significantly greater hip extension and knee flexion angles during the early swing phases of sprinting, resulting in anterior pelvic tilt. Pelvic coordination has been highlighted as a possible risk factor for HSI incidence (Schuermans, Tiggelen, et al., 2017), specific kinematic changes in pelvic co-ordination could be influencing forces experienced by the hamstrings leading to injury. Recently, Alt, et al. (2020) highlighted that concentric hip mechanics are significantly related with the mechanics at the knee, although the authors only observed the terminal swing phase of sprinting (the most high risk phase of running), nevertheless it highlights that the changes observed within the present study during the early phases could be influencing the later phases of the swing phase at the knee and hip.

Within the present study knee joint kinematics were also found to be altered, with high-risk group going through greater knee flexion, extension and change in knee angular velocity. The characteristics that the high-risk group demonstrate could be associated with "back side" mechanics, where their foot is actively pulled towards the hip through greater knee flexion, with subsequent greater knee extension at greater knee angular velocities during the terminal swing phase. Under fatigued conditions, there is a reduction in peak hamstring length as a potential protective mechanism to prevent possible overstretching and subsequent muscle damage, however, Small, et al. (2009) presented similar findings to the present study, where the shank underwent greater segment velocities during the terminal swing phase, resulting in the lower leg being "whipped" through. Thus, creating additional strain on a system which is weaker, either eccentrically weaker as within the present study or through fatigue, increasing HSI risk (Small, et al., 2009). In addition, Alt, et al. (2020) found that eccentric hamstring strength was significantly related to knee joint kinetic parameters observed during the late swing phase of maximal sprinting. During the terminal swing-phase the athlete's shank weight would influence the knee angular velocity and the subsequent momentum. Therefore, a peak eccentric hamstring strength value that is relative of an athlete's body weight (i.e., relative peak torque), could discriminate more effectively for knee joint kinematics. This highlights those high-risk individuals (decreased relative BF_{LH} FL and relative eccentric peak torque) had a greater knee flexion, followed by greater knee extension at higher change in knee angular velocities therefore the high-risk individuals would be required to dissipate a greater amount of kinetic energy, indicating a decreased ability to control the extending limb particularly through the descending length-tension curve.

The results within the present study are consistent with what has been previously reported for individuals who have sustained a previous HSI (Brughelli, et al., 2011). Within the study conducted by Brughelli and colleagues (2011) a previously injured hamstring produced a significantly lower horizontal force when running at 80% of maximal running velocity. Moreover, consistent with present findings (for the high-risk group), the injured cohort had a shorter contact time, impulse and leg stiffness than the non-injured controls, albeit non-significant difference (Brughelli, et al., 2011). There results potentially indicate, that there must be a change in joint action velocity which could be a motor adaptation, as a method to compensate for a reduced force generating capacity. Although contrastingly, no significant differences were observed between peak hip and knee angular velocities between individuals with a history of previous HSI and no history at the same running velocity of 80% of maximal running velocity (Lee, et al., 2009). This difference in findings between the two studies (Brughelli et al.(2011) vs. Lee et al.(2009)) could be a result of different time periods following injury occurrence (< 2 years Vs < 3 years). Furthermore, the present study utilized a mixed team-sport population; while Brughelli et al.(2011) utilized a single team-sport population (AFL) and Lee et al.(2009) used mixed sport population including team-sport, track and field and endurance sports, which all may display slightly altered kinematics regardless of injury history (Brughelli, et al., 2011; Lee, et al., 2009).

The present study is not without its limitation, firstly, the top speed of the treadmill was only 16 $km \cdot hr^{-1}$, which by team sport standards is nowhere near high speed running velocities, potentially categorized into striding within team sports (Sweeting, Cormack, Morgan, & Aughey, 2017). Although the treadmill was not able to achieve maximal speeds, there was a nearly perfect positive correlation with the running velocity and magnitude of differences between the two groups (Appendix Three). Essentially, as running velocity increased from 8 km·hr^{-1} to 16 km·hr^{-1}, the difference in running kinematics and activation between the high- and low-risk groups also increased, potentially indicating that at higher velocities, closer to maximal speeds, the differences between groups could have been exaggerated. Secondly, it has suggested previously that there are differences in running biomechanics observed between treadmill and over-ground running, despite the research being conflicting. Van Hooren et al. (2020) performed a systematic review and meta-analyses across studies, identifying that spatiotemporal, kinematic, kinetic, muscle activity and muscle-tendon outcome measures are largely comparable between treadmill and overground running. Although a number of sagittal plane outcome measures differed between treadmill and overground running, including, sagittal foot-ground angle at TD, knee flexion at TD, knee flexion ROM during the stance phase and vertical displacement of the pelvis (Van Hooren, Fuller, et al., 2020). Furthermore, hip angle at TO was greater when running on a treadmill when compared to track-based running (Van Hooren, Fuller, et al., 2020). The findings of the systematic review and meta-analyses could indicate that some of the observations in sagittal plane running kinematics may at the very least be exaggerated when running on a treadmill (Van Hooren, Fuller, et al., 2020). However, this does not take away from the differences observed with the present study, with many of the observations being comparable to overground running (Van Hooren, Fuller, et al., 2020), with only a few outcome measures that may be magnified, such as peak hip extension, it would be expected that the increase in peak hip extension would be similar for both groups meaning a difference would still be observable regardless of running surface.

A potential consideration which could have improved the present study, could have been to have athletes sprint through the 3D motion capture area, and thus achieving a near maximal running velocity. In hindsight, there could have been a more appropriate method of analysing the effect of the modifiable risk factors of HSI on a high-risk task (i.e., sprint running). However, there are issues with this method, including individual acceleration profiles, where different phases of running would have been assessed, additionally this would also lead to non-standardised running velocities. Moreover, there was some concern over 3D marker, EMG electrode, and participant safety if athletes were maximally sprint running, particularly with the short run off in the University of Salford biomechanics laboratory.

5.1.4.6 Electromyography differences

A key finding from the current study is that across all running speeds, those considered to be at an elevated risk of HSI, demonstrated greater relative activation of the BF_{LH} at all running velocities with a more symmetrical intra-muscular patterning between lateral and medial components. This indicates that regardless of running velocity, the BF_{LH} activates to a greater relative intensity within high-risk individuals (e.g., weaker and shorter BF_{LH} FL), which could potentially be a causative factor, for HSI particularly if this trend continued to increase at greater running velocities, i.e., >16 km·hr^{-1}.

The literature is mixed upon the effect of previous HSI on hamstring EMG activation, with no significant difference in BF_{LH} activity between healthy and previously injured limbs when running at 60-, 70-, 80-, 90- and 100% of maximum (Silder, Thelen, et al., 2010). Whereas during prone hip extension a significant increase was observed in lateral hamstring activation (i.e. BF_{LH}) between previously and non-previously injured individuals (Emami, et al., 2014). However, there are several methodological dissimilarities to the present study, firstly, Silder et al. (2010) compared within participants, i.e. previously injured limb vs non-previously injured limb, in lieu of a "healthy" non-previously injured cohort. This may have influenced the results, as individuals could have developed compensatory activation patterns and strategies in response to the injury. Furthermore, the current study only tested a maximum running velocity of 16 $km \cdot hr^{-1}$, which would equate to approximately 60% of maximum running velocity achieved during normalization, in contrast to the higher velocity running used by Silder et al. (2010).

The relative activation of the ST was similar between groups across all running velocities, furthermore, it was consistent with previous literature, where the peak relative neuromuscular contribution of the ST was greater across all running velocities in comparison to the BF_{LH} (Higashihara, et al., 2018). Although the relative contribution between both lateral and medial components does highlight that that at the higher running velocities, high risk individuals do have a greater neuromuscular contribution of the BF_{LH}. Recent investigations sought to determine if the BF_{LH} and ST are teammates or competitors? That's the question Schuermans et al. (2014) attempted to answer regarding the activation patterns of the BF_{LH} and ST muscles during exercise. Using functional MRI to map the intra- and intermuscular activation patterns during a knee extension and flexion task, they observed that a complex synergistic patterning between the BF_{LH} and ST, although the ST was more predominantly activated during the eccentric action within footballers with and without a history of HSI. Further to this they identified a more symmetrical activation pattern between BF_{LH} and ST, within the injury group compared to the control group during this eccentric action. This is consistent with the results of the present study, where the relative contribution of the BF_{LH} was more evenly matched to the ST, for the high-risk groups compared to the low-risk groups at the higher running velocities. However, Schuermans and colleagues (2014) used a knee extension-flexion task that mimicked the hamstring mechanics within running - although the task that was performed could not have the potential to replicate the loading that occurs during maximal velocity running, due to limited load and velocity. Furthermore, there is a large difference in the methods used to assess muscle activation between studies, Schuermans et al.(2014) assessed muscle activation from the transverse relaxation time (T2) pre- and post-exercise in contrast to the surface EMG within the present study. Between EMG and T2 estimations of muscle activation, significant and meaningful associations have only been observed for the ST during eccentric exercise (Kubota et al., 2009), supporting the findings of the present study. Additionally, Emami, Arab and Ghamkar (2014) demonstrated that during a prone hip extension task, there was no significant difference within medial hamstring activation (i.e. ST and SM), between previously and non-previously injured individuals, although again this is across a different muscle action and movement - therefore it is unsurprising that the results differ to those found within the present study.

The focus of the present thesis was on hamstring including assessment, performance, and training, with an underlying objective to observe if the modifiable risk factors of HSI influence the hamstring functioning during running. However, as high-speed running is a global multi-joint task it may have been useful to examine a more extensive range of lower limb muscles, such as the adductor magnus, glutes and gastrocnemius, to get an idea about the supporting roles of these muscles and the inter-muscular coordination that occurs during the gait cycle, with many of these muscles having multiple functions (Avrillon, Guilhem, Barthelemy, & Hug, 2018; Besier, Lloyd, & Ackland, 2003; Camic, Kovacs, Enquist, McLain, & Hill, 2015; Ekstrom, et al., 2007; Emami, et al., 2014; Li, Landin, Grodesky, & Myers, 2002). Therefore, future work should consider looking across these muscles and how they interact and may potentially even compensate for a weakness within the hamstrings, which could even be leading to secondary injuries (e.g., gastrocnemius or adductor strains). Furthermore, with the establishment and availability of more advanced EMG technology, such as HDEMG, where a linear array of electrodes allows for the identification of region specific activation profiles, this information would further benefit research providing more exploratory detail around the gait cycle and potential differences regional activation differences which could play a role in HSI (Blandford, et al., 2018; Fyfe, et al., 2013; Hegyi, Csala, Peter, Finni, & Cronin, 2019; Hegyi, Gonçalves, Finni, & Cronin, 2019; Hegyi, Peter, Finni, & Cronin, 2018).

5.1.5 Conclusion

The information within the present study (Part A), highlights a novel finding with regards to the effect of the modifiable risk factors of HSI (eccentric hamstring strength and BF_{LH} FL), on dynamic high-risk task performance. It was observed that individuals possess a reduced relative eccentric hamstring strength and relative BF_{LH} FL, performed sub-maximal running with altered kinematic and EMG patterns. Specifically, there were significant differences observed for knee and hip kinematics and BF activation were greater for the perceived high-risk group. The differences highlighted between groups could, therefore, be an influencing factor in HSI occurrence. Peak change in knee angular velocity was observed to occur within the late swing phase, which has been hypothesised to be the phase where HSIs are likely to occur (Heiderscheit, et al., 2005). If a greater change in knee angular velocity is achieved during the terminal swing phase (knee extension), which coincides with reduced eccentric hamstring strength and BF_{LH} FL; this could increase the inertial loads experienced within the hamstrings across the descending length-tension relationship, ultimately increasing damage susceptibility. Although, this is purely conjecture as inertial loads were not observed as part of the present study. Furthermore, the changes observed across both the knee and hip joints could be a result of sub-optimal pelvic control, with anterior pelvic tilt being highlighted as a potential high-risk action occurring in those who sustain a future HSI and under fatigued conditions (Schuermans, Danneels, et al., 2017; Schuermans, Tiggelen, et al., 2017; Small, et al., 2009). The present study demonstrates that even prior to injury – running mechanics and activation profiles are different as a result of the relative eccentric hamstring strength and relative BF fascicle length. Although as recently suggested more prospective studies are required examining running mechanisms (Kenneally-Dabrowski, Brown, Lai, et al., 2019; Kenneally-Dabrowski, Brown, Warmenhoven, et al., 2019),

5.1.6 Linking Paragraph

The present study (Chapter 5-Part A) highlighted two key observations; firstly, stronger associations were observed between relative measures of BF_{LH} FL and isokinetic strength measures. Using similar formatting as Timmins, Bourne, et al. (2016) where a quadrant of doom was designed enabling identification of those who could be of high or low risk, being either "short and weak" or "long and strong", respectively. Participants who were classified as "short and weak", performed a running task with what could be described as unfavourable lower limb kinematics and hamstring activation patterns, potentially elevating the risk of HSI incidence. However, only the peak characteristics were observed, which only tells a small portion of the story with a considerable portion of the gait cycle excluded from analyses. It also overlooks the terminal swing phase, which could be described as the most hazardous portion gait cycle where even a small difference could be a potential cause of a HSI event. Therefore, the entire waveform does require observation to identify where within the gait cycle difference are occurring, how the measures could be interacting and if they could be related to the proposed mechanisms of future HSI occurrence during running based tasks (Askling, Karlsson, & Thorstensson, 2003b; Chumanov, et al., 2011; Heiderscheit, et al., 2005; Higashihara, et al., 2015, 2018; Van Hooren and Bosch, 2017a, 2017b).

5.2 Part B – Effect of the modifiable risk factors of Hamstring strain injury upon Waveform kinematic and activation during running

5.2.1 Introduction

The high rate of HSI occurrence has been identified being significantly influenced by eccentric hamstring strength and BF_{LH} FL, i.e., the modifiable risk factors (Bourne, et al., 2015; Opar, et al., 2015; Opar, et al., 2012; Ruddy, Shield, et al., 2018; Timmins, Bourne, et al., 2016). Moreover the inclusion of an eccentric training stimulus can result in rapid increases in both hamstring strength and BF_{LH} FL (Bourne, et al., 2018), being effective in reducing HSI occurrence (van Dyk, et al., 2019), as well as improvements in the performance of athletic tasks such as running and jumping (Chu, et al., 2017; Freeman, et al., 2019; Ishoi, et al., 2018; Krommes, et al., 2017). This indicates that athletic performance of tasks could be associated with changes in hamstring strength and BF_{LH} architecture. A recent pre-print could add weight to this suggestion where the authors observed that eccentric hamstring strength was significantly associated with absolute sprint performance, in addition to the kinematics and kinetics of the knee during the late swing phase (Alt, et al., 2020; Alt, et al., 2021).

Alterations in running kinematics and kinetics, kicking kinematics, muscle activation patterns and lengthening muscle tissue mechanics have been observed within individuals with a history of HSI (Brughelli, et al., 2011; Emami, et al., 2014; Lee, et al., 2009; Lord, Blazevich, et al., 2018; Navandar, et al., 2017; Silder, Thelen, et al., 2010). However, a key limitation of all these studies, is that they all have taken a retrospective look on hamstring and athletic performance following HSI (Brughelli, et al., 2011; Emami, et al., 2014; Lee, et al., 2009; Lord, Blazevich, et al., 2018; Navandar, et al., 2017; Silder, Thelen, et al., 2010), whereby the kinetic, kinematic, activation and mechanical changes could be a result of motor adaptation to protect the system from further injury (Hodges and Tucker, 2011). Additionally, this research has focused upon peak differences between individuals, ultimately this limits the usefulness of this research as a peak moment can occur for a single 100th of a second, which may not be indicative of what could be occurring across the whole gait. Therefore, research should aim to observe significant and meaningful differences that could be present between groups across the entire gait cycle. One potential method to achieve this could include the use of statistical parametric mapping (SPM), which retrieves results on statistical inference, based upon the Random Field Theory.

Conceptually, the SPM analysis process is similar to the calculation and interpretation of a scalar two-sample *t*-test (Robinson, Vanrenterghem, & Pataky, 2015). However, it provides the ability to statistically analyse whole movements that are represented by relevant time-series, referred to as waveforms (Warmenhoven et al., 2018). Differences within individual waveforms (i.e. waveform shape), has been termed as an individual movement signature (Warmenhoven, et al., 2018). The potential of individual signatures within different biomechanical waveforms has key implications within sport and exercise, with ability to statistically differentiate markers in performance (kinematic or muscular activation) between groups (Warmenhoven, et al., 2018). The information that SPM analyses of kinematic and EMG waveforms could provide, may highlight differences within the individual signatures, between those who are "strong and long" and "short and weak"
The results from part A, demonstrated significant and meaningful differences between peak kinematic and EMG measures. However, the peak kinematic or EMG measure occurs for ≈0.015 s, therefore, does not provide the detail of what is occurring across the whole gait cycle between groups, which may arguably be more important with regards to HSI mechanisms. Therefore, a novel aspect of the part B is the addition of waveform characterization and statistical analyses, where differences were assessed across the whole gait cycle.

5.2.2 Aims and Hypothesis

The primary aim of part B of this study was to observe if there were any observable differences in running kinematics and hamstring activation patterning across the entire gait cycle (waveform), between individuals who are perceived to be at a high or low risk of HSI occurrence as according to the modifiable risk factors of HSI; hamstring strength and BF_{LH} architecture. It was hypothesized that a significant difference in running kinematics and activation would exist between high and low risk groups.

5.2.3 Methods

A full description of the methodological procedures can be found within Chapter 5 – Part A. Waveforms within the present study were only observed at 16 km·hr^{-1}, as within Chapter 5 – Part A, the largest differences were observed at the greatest running velocity (16 km·hr^{-1}).

5.2.3.1 Statistical analysis

5.2.3.2 Kinematic and Electromyography statistical approach

Mean kinematic and EMG waveforms (knee and hip joint angles, knee joint angular-velocity changes, BF_{LH} MTU length and BF:ST EMG) were temporally normalized from 0% to 100% of the gait cycle (TD to subsequent TD) for both high (*n* = 8) and low risk groups (*n* = 8), at 16 km·hr^{-1}. A coefficient of multiple correlations (CMC) with 95% CI was performed to analyse the between stride reliability between waveforms by comparing shape, timing and amplitude (Van Hooren, Teratslas, et al., 2020). The CMC values were be interpreted based on the 95% CIs as (<0.50) poor, (0.5-0.74) moderate, (0.75-0.90) good and (>0.90) excellent (Koo and Li, 2016).

Differences between group time normalized traces were compared across multiple methods within a custom Microsoft Excel spreadsheet. Firstly, likely differences between groups were determined by plotting the time normalized curves for each group along with the corresponding upper and lower 95% CIs to create upper and lower control limits, identifying non-overlapping areas, indicating likely differences. Secondly, differences between groups were compared using one-dimensional two-sample *t* tests, where *t* tests were applied to each of the 101 nodes resulting in a *t* curve. If the *t* curve exceeded a statistical *t* threshold, there was deemed to be a significant difference. Finally, the magnitude of differences between groups were determined by Cohen's *d* effect sizes plotted across 101 nodes. Defined threshold values of values of -0.35 to 0 and 0 to 0.35 as trivial, -0.35 to -0.80 and 0.35 to 0.80 as small, -0.80 to -1.50 and 0.80 to 1.5 for moderate and > -1.5 and > 1.5 as large differences (Rhea, 2004).

5.2.4 Results

All data was determined as being normally distributed (p >0.05).

5.2.4.1 Kinematic and EMG reliability data

5.2.4.2 Between-stride reliability

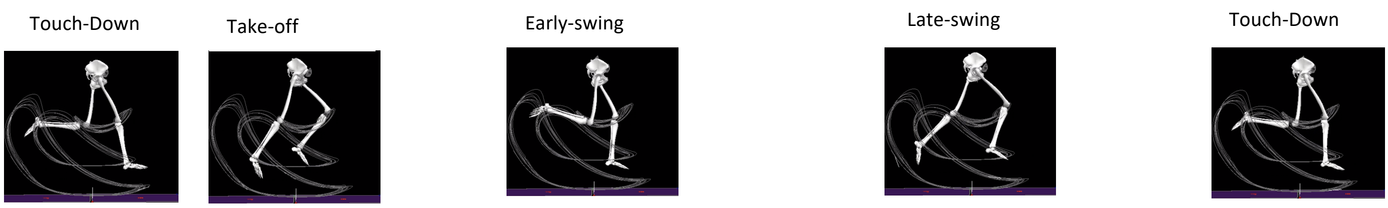
Between stride waveform reliability (CMC), found moderate to excellent relative reliability. Kinematic waveforms including, hip angle (CMC (95% CI) = 0.931 (0.874 – 0.985)) and knee angle (CMC (95% CI) = 0.940 (0.890 – 0.990)) were found to have good reliability, whereas change in knee angular velocity had excellent reliability (CMC (95% CI) = 0.974 (0.943 – 0.996)). A moderate-excellent level of relative reliability was found for BF:ST activation ratio (CMC (95% CI) = 0.881 (0.736 – 0.999)). This highlights that between strides, waveforms were of comparable shape, timing and amplitude.

5.2.4.3 Waveform kinematic and EMG characteristics between groups

Consistent significant differences were observed for both left and right limbs (within part A), therefore, the waveform differences for a single (right) limb are presented here.

5.2.4.4 Hip Angle Waveform

Significant and moderate differences were observed throughout the waveform (p < 0.05, d = 0.80 – 1.50), with the greatest differences observed in the early and late phases of the swing phase (Figure 5-6).



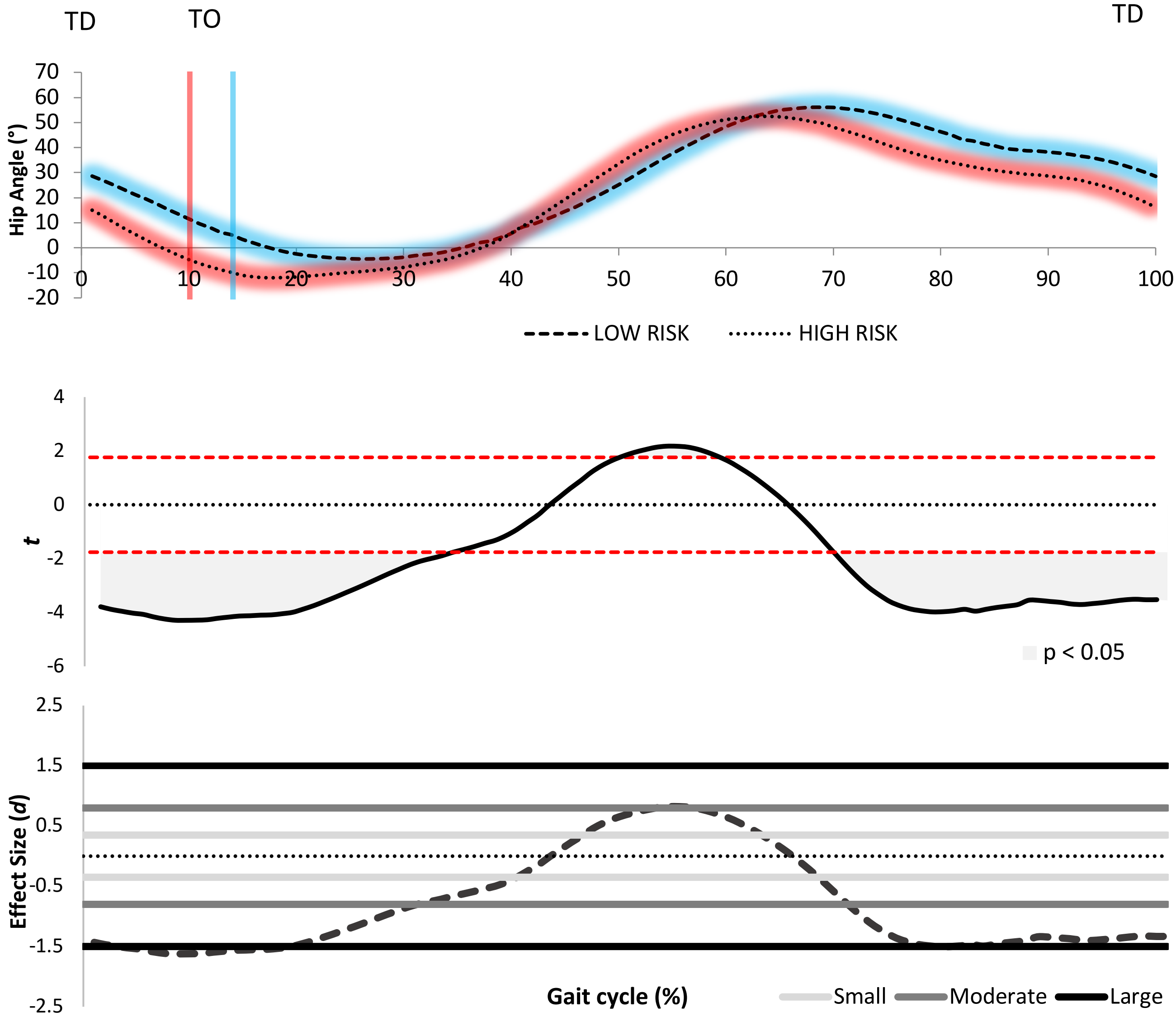


Figure 5-7 Hip angle waveform with 95% confidence intervals and statistical differences signatures (t statistic and Cohen's *d). TD = Touch down, TO = Take off.*

5.2.4.5 Knee Angle Waveform

Significant and moderate differences were observed between groups throughout the waveform ($p < 0.05$, $d = 0.80 – 1.50$). The greatest differences were observed throughout the swing phase (Figure 5-7).

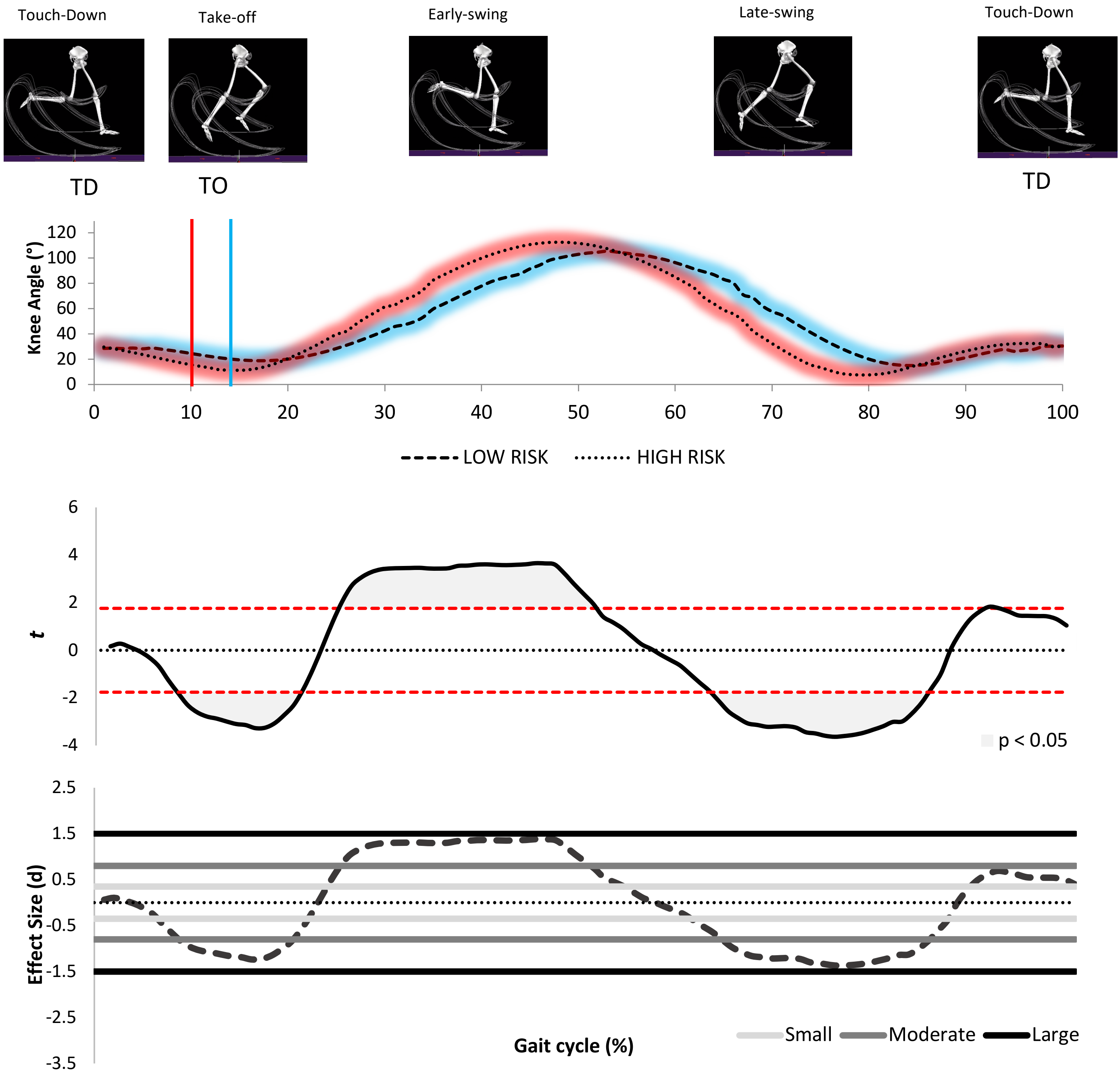


Figure 5-8 Knee angle waveform with 95% confidence intervals and statistical differences signatures (t statistic and Cohen's d). TD = Touch down, TO = Take off.

5.2.4.6 Biceps femoris long head muscle tendon unit length

The BF_{LH} MTU waveforms, demonstrated significant and moderate to large difference between groups ($p < 0.05$, $d = > 1.50$), with differences observed across the gait cycle (Figure 5-8).

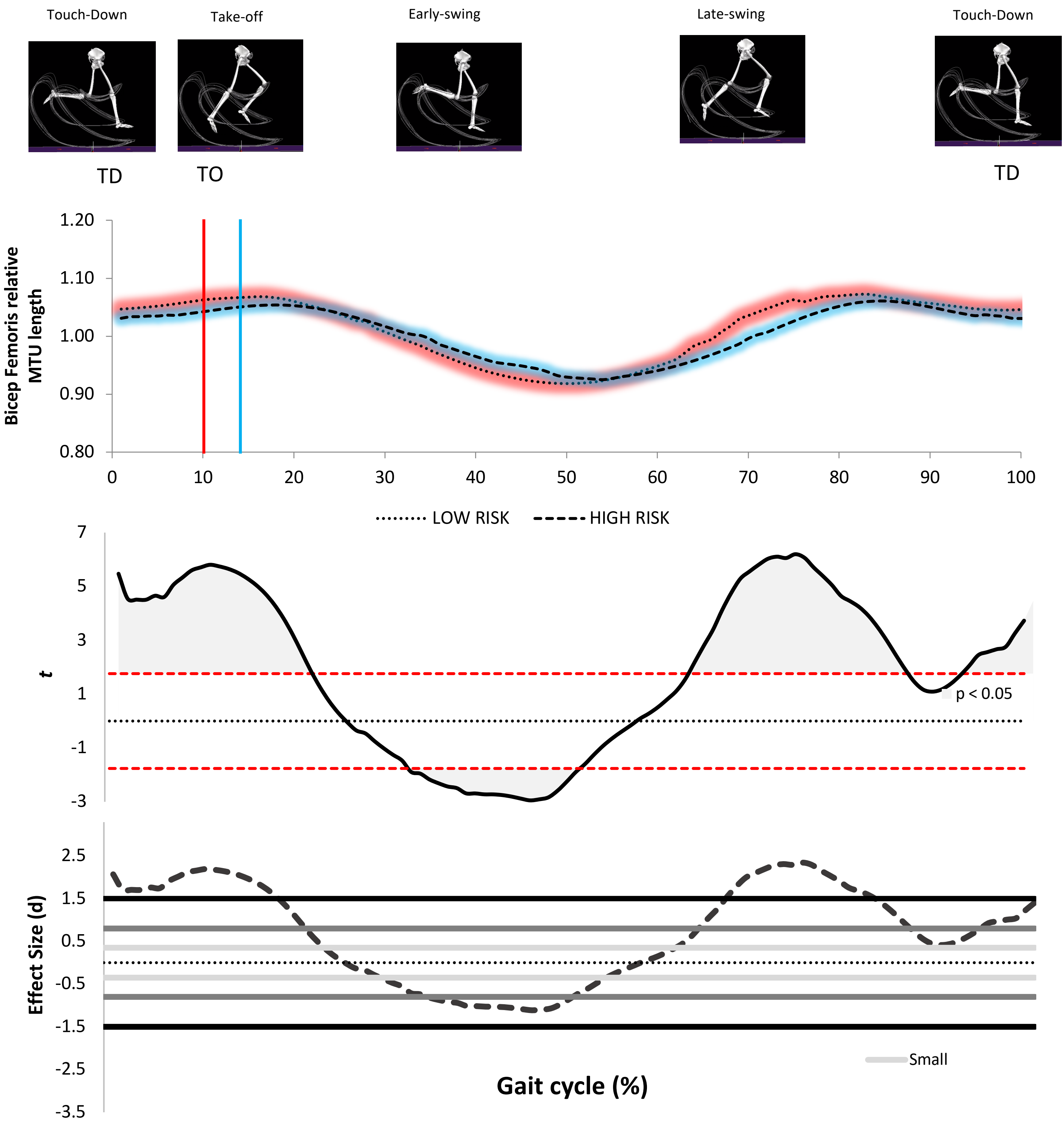


Figure 5-9 Bicep femoris muscle-tendon unit length waveform with 95% confidence intervals and statistical differences signatures (t statistic and Cohen's d). TD = Touch down, TO = Take off.

5.2.4.7 Change in Knee Angular Velocity

The change in knee angular velocity waveforms, demonstrated significant and large difference between groups ($p < 0.05$, d = > 1.50), with the greatest differences observed during the mid-swing phase (Figure 5-9).

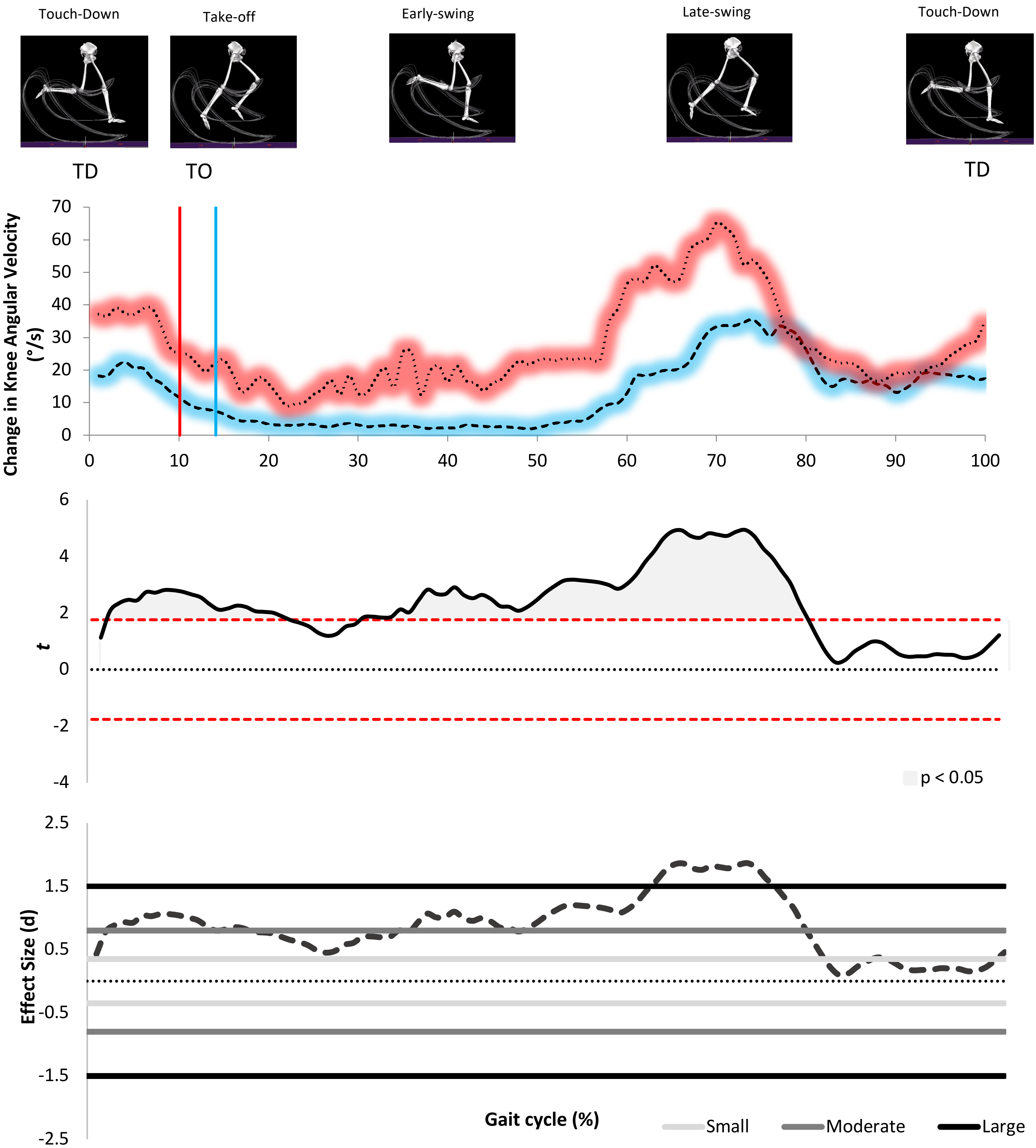


Figure 5-10 Change in knee angular velocity waveform with 95% confidence intervals and statistical differences signatures (t statistic and Cohen's d). TD = Touch down, TO = Take off.

5.2.4.8 *Lateral to medial hamstring activation ratio*

The lateral to medial hamstring activation ratio waveforms, demonstrated a significant and large difference between groups (*p* < 0.05, *d* = > 1.50) in lateral hamstring emphasis during the late swing phase (Figure 5-10).

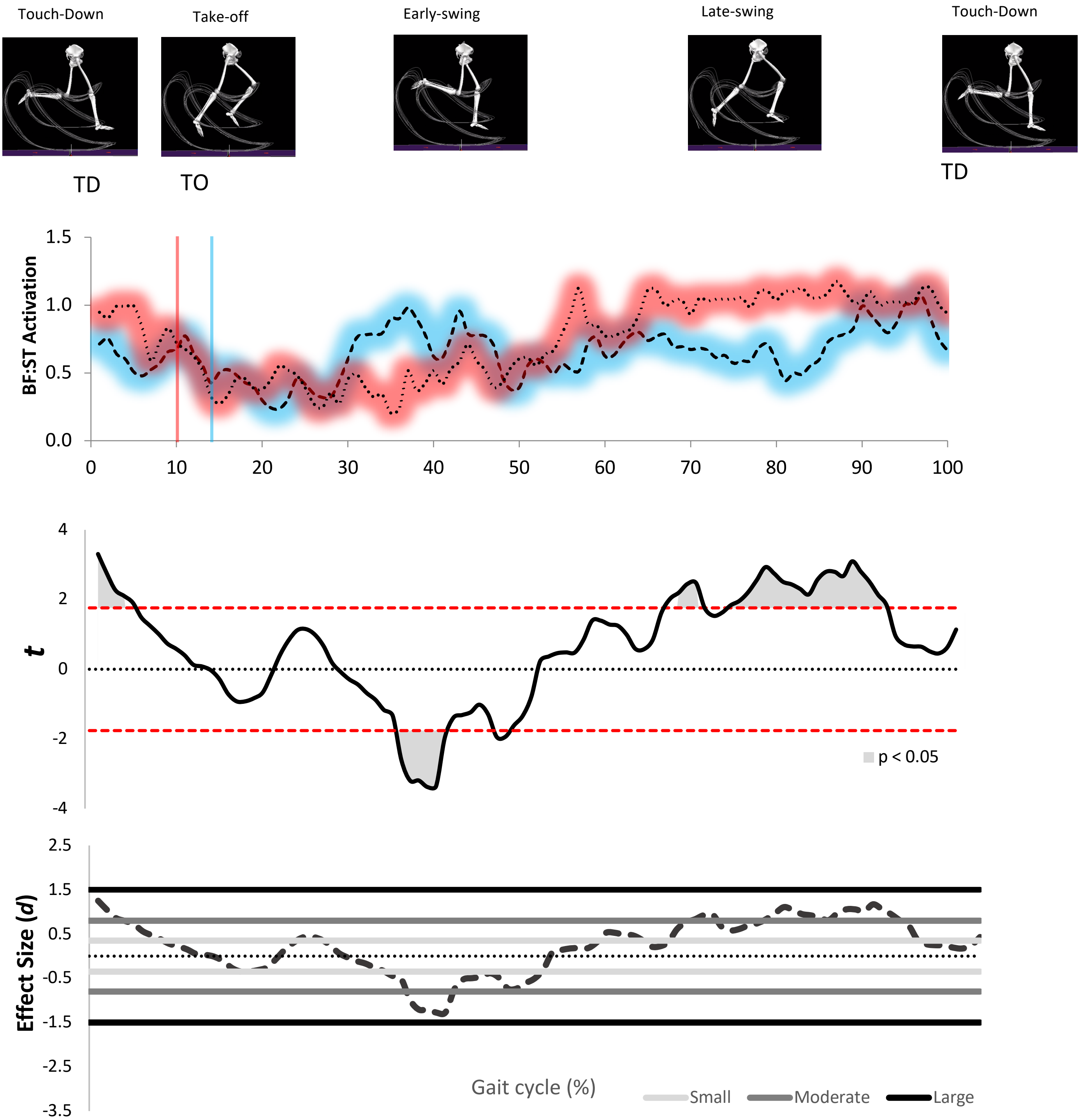


Figure 5-11 Lateral to medial hamstring activation waveform with 95% confidence intervals and statistical differences signatures (t statistic and Cohen's d). TD = Touch down, TO = Take off.

5.2.5 Discussion

5.2.5.1 Kinematic signature differences

Each of the individual biomechanical signatures demonstrated significant and meaningful differences between the high- and low-risk groups, across phases of the waveforms. Additionally, the differences observed were consistent between limbs, with greater running velocities resulting in more pronounced differences in the biomechanical signatures. The kinematic signature waveforms for both high- and low-risk groups highlight key trends in hip, knee ROM and BF_{LH} MTU length (Figure 5-6, 5-7 and 5-8) and change in knee angular velocity (Figure 5-9). If the observed, concurrent increase in kinematic differences and running velocity continued up to maximal speed, may potentially be elevating injury risk.

Across the kinematic signatures of the hip, knee BF_{LH} MTU length, significant and meaningful differences were observed across multiple phases in the gait cycle. For hip ROM, significant and meaningful differences were observed at the beginning and end of the gait cycle, i.e., stance and early swing and late swing prior to touch down. During the stance and early swing phase, the high-risk group demonstrated a greater magnitude and duration of hip extension. The data within the present study contrasts that reported by Lee et al.(2009), where no identifiable differences within hip ROM could be inferred across the gait cycle. Although, Lee et al.(2009) only presented mean waveforms with confidence interval shading and made no statistical inference between the waveform signatures (injured vs control), using only peak values for statistical inference. However, as referenced within Part A, Small, et al. (2009) found that under-fatigued conditions, during sprinting soccer player present greater hip extension and knee flexion, this matches what was observed in the present study. Under fatigue a muscle has a reduced force generating capacity, i.e., its weaker, this could explain the findings of the present study, where those at high-risk (i.e., eccentrically weaker and reduced BF_{LH} FL) had similar kinematics, with greater hip extension and knee flexion within the first half of the swing phase.

As suggested within part A of the present study, knee ROM between groups was meaningfully different. The waveform signature identified three phases across the gait cycle, which were significantly and meaningfully different between groups. The differences highlighted occurred at take-off, early swing and late swing phases, with no consistency across the waveform. The data demonstrates that the high-risk group are going a considerably greater ROM at knee, which could be indicative of "backside mechanics" with altered heel recovery, especially when hip kinematics are also considered. With significantly greater knee extension at take-off, increased knee flexion within the early swing phase and increased knee extension during the late swing phase. The greater ROM observed for the high-risk group, can therefore explain the increased change in knee angular velocity across the entire gait cycle. Furthermore, the greatest differences in change in knee angular velocity occurred later within the swing phase, as the knee reaches peak extension, this is consistent with Small, et al. (2009), where the shank underwent greater segment velocities during the terminal swing phase, resulting in the lower leg being "whipped" through as it reaches peak knee extension. The greater velocity suggests there is a greater amount of kinetic energy that requires absorbing, occurring within the terminal swing phase, which would have implications upon injury risk and potential muscle damage, as reductions in eccentric hamstring strength would impair their ability to decelerate the shank (Small, et al., 2009). However, Alt, et al. (2020) found no significant relationship between kinematic measures of eccentric hamstring strength (angle of peak moment and angle of peak power) and knee joint angular velocities in the late swing phase. However, the authors did look to identify if any potential relationship exists between kinetic measures of isokinetic eccentric hamstring strength (peak moment, contractional work, peak power and mean power) and knee joint angular velocities, which help to explain the observations within the present study.

This if the first study to attempt to observe kinematic movement signature differences between groups that were stratified based on relative eccentric hamstring strength and BF FL. In accordance with Part A, the combination of relative BF_{LH} FL and relative eccentric peak torque appears to have a meaningful impact upon global running performance, with consistent differences observed between limbs. The additional information attained from Part B highlights a potential difference in running strategy (e.g., heel recovery), which could be predisposing individuals to an elevated risk of HSI. Furthermore, the data presents unique information on the potential for the kinematic measures to be inter linked, whereby an effect on one influence the other two, although it is impossible to determine what may be the "cause" or the "consequence" of each difference without further, more in-depth investigation. However, future research should look at a wider range of kinematic and kinetic variables that may provide a deeper understanding of some of the outcome measures found within the present study. Kinematic measures such as pelvic motion (including anterior-posterior pelvic tilt), moment of inertia and angular momentum of the swing leg, and centre of mass of the foot velocity throughout the gait cycle. Investigating these outcome measures may provide further support as to why those who would be considered "short and weak", present greater angular velocities and activation profiles, as the hamstrings are unable to decelerate the shank segment (i.e., greater centre of mass foot velocity and greater momentum) or a definitive answer on the role of the modifiable risk factors in hip extension in conjunction with pelvic tilt.

5.2.5.2 Electromyography differences

Similar to the peak EMG differences observed within Chapter 5 Part A, there was a significant and meaningful increase in relative BF activation and thus the relative contribution of the BF when compared to the ST (BF:ST activation ratio) (Figure 5-10). The present data set highlights that the observed differences between high and low-risk groups activation patterns occurs late within the swing phase prior to the subsequent touch-down. This finding could explain the elevated injury risk to the BF_{LH} during higher velocity running within the terminal swing phase. Furthermore, there was a concurrent increase in activation of the BF_{LH} as running velocity increased (Chapter 5 Part A), which would be expected to continue at greater velocities.

An increased activation of the BF_{LH} during the late swing phase, where the BF_{LH} is undergoing a lengthening action reaching its greatest strain (>110% of anatomical resting length) (Heiderscheit, et al., 2005; Van Hooren and Bosch, 2017a, 2018), could be elevating strain potential as the hamstrings attempt to actively resist the extending knee. The increased potential of a strain injury occurrence relates to the "popping sarcomere" theory (Morgan and Proske, 2004). As the hamstrings are applying a resistive force as they undergo the greatest strain (i.e. late swing phase), which leads to sub-optimal overlap of the sliding filaments, there is the potential for stretch induced muscle damage (Morgan and Proske, 2004). Although the cause of the resulting injury is still mainly conjecture within the literature, as researchers cite either a single catastrophic event or repeated microscopic areas of damage, weakening the muscle to an extent where a catastrophic event occurs (Morgan and Proske, 2004; Opar, et al., 2012). The combination of the high magnitude of strain and increased activation demands, during the terminal swing phase for the high-risk group within the present study, could be indicative of an elevated potential for stretch induced muscle damage.

Previous research has demonstrated that a HSI event, impacts upon the neuromuscular activation patterns, although generally these have focused upon peak values (Emami, et al., 2014; Opar, Williams, et al., 2013b; Schuermans, et al., 2014). Each study has observed differences in activation patterns across athletic tasks, including concentric and eccentric isokinetics (Opar, Williams, et al., 2013b), hip extension isometrics (Emami, et al., 2014), and simulated running demands (Schuermans, et al., 2014). However, the differences observed have not been consistent across the literature, potentially highlighting that the neuromuscular differences as an effect of injury or in the case of the present study pre-injury could be task specific. Although it should be emphasised that the differences found in previous literature could also be an effect of individual injury characteristics i.e., time since HSI, location, grade etc, which is not commonly reported due to limited diagnosis availability.

5.2.6 Conclusions

The waveform differences identified within the present study, highlight key phases of the gait cycle where differences between those who have a greater relative eccentric strength and relative BF_{LH} FL. Key differences typically occurred within the late swing phase, with the emphasis shift to BF_{LH} activation over ST for the weaker and shorter group (high-risk group), while concurrently a greater change in knee angular velocity was occurring. There were differences in kinematics observed across the gait, highlighting potential varying running strategies, including a shift to "backside mechanics" and heel recovery differences. Backside mechanics may potentially have some advantages to sport performance, as long it does not hamper sprint performance, specifically certain scenarios which may encountered within sports, such as contact team sports, it might be advantageous to avoid contact or break tackles via variability and overall control of the system to be able to change strategy (i.e., gait cycle) within a performance-injury conflict. This however highlights the need for high levels of eccentric hamstring strength and BF_{LH} FL for optimal preparedness, if an athlete must move through low- to high-risk strategies to achieve optimal sporting performance, then they require high levels of preparedness to minimise the risk of HSI incidence. Additionally, the differences highlighted within chapter 5 could help to understand the observed within EMG – due to the effect of muscle length and lengthening velocities the hamstring would under-go as an effect of different mechanics (Schuermans, Danneels, et al., 2017; Schuermans, Tiggelen, et al., 2017; Small, et al., 2009). In accordance with the previous conclusion (Part A), prospective differences in running mechanisms and activation are altered across the gait cycle, as an effect of the modifiable risk factors of HSI.

It should be noted that the present data set does not include “true” high speed running, achieving only striding velocities of 16 km·hr^{-1} (Sweeting, et al., 2017). Although, with consistent differences observed between limbs and strong linear increases in the observed differences as an effect of speed, it would be hypothesised that at greater to near maximal velocities the differences could increase, between those who “strong and long” and “short and weak”. However, to confirm this hypothesis further investigation is required, with interest in team sports and sprint athletes where differences in sprint kinetics have been identified.

5.2.7 Linking paragraph

Chapter 5 highlighted novel potential mechanisms of HSI incidence, specifically how the two modifiable risk factors (eccentric hamstring strength and BF_{LH} FL), influence the kinematic and neuromuscular performance of a dynamic task (i.e., running). Therefore, effective HSI prevention practices such as the NHE, that decrease the risk of HSIs, by positive increases in eccentric hamstring strength and BF_{LH} FL should be the primary consideration. Furthermore, optimising the NHE via an appropriate and standardised regression or progression could positively influence athlete buy-in and compliance, which is a key factor in HSI prevention effectiveness (Chapter 2-5).

To date, regressions to the NHE have included harness and band assisted NHEs (Alt, et al., 2018; Alt, et al., 2021; Matthews, et al., 2017), whilst progressions include the addition of load. Both have shown positive beneficial effects, however, assisted variants are impractical to work in strength and conditioning team sport environments or are unable to be standardised. Therefore, alternatives should be explored that have the same potential to lead to positive benefits in eccentric hamstring strength and BF_{LH} FL, while being practical and able to be standardised.

6 Study 4 – Kinematic, neuromuscular and bicep femoris *iv vivo* mechanics of the Nordic hamstring exercise and variations of the Nordic hamstring exercise

6.1 Introduction

The hamstrings muscle structure including FL, is extremely pliable in response to different training stimuli, which is extremely relevant in the reduction of HSI risk (Bourne, et al., 2018). Supramaximal eccentric exercises, such as the NHE, have been shown to increase BF_{LH} FL (Chapter 2.6). This is in contrast to quasi-isometric (e.g., razor curls) and short muscle length conventional hamstring training (e.g., lying leg curls), where there was a no change in and a decrease in BF_{LH} FL, respectively (Bourne, et al., 2018; Duhig, et al., 2019; Pollard, et al., 2019). However, to date, research is limited in quantifying the muscle fascicle dynamics during any hamstring resistance exercises. This information may aid researchers and practitioners in explaining why preferential adaptations (i.e., increased BF_{LH} FL), may occur when utilising the NHE.

Cattaneo (2018), recently explored the muscle fascicle dynamics of eccentric biased, hamstring rehabilitation exercises utilising dynamic US. The exercises that were assessed encompassed the Askling rehabilitation exercises (Askling, et al., 2014), which are three body weight exercises (glider, slider and extender). These exercises are eccentric are unloaded or low-load hamstring lengthening actions with minimal to no negative work being performed (Askling, et al., 2014). However, due to poor US image or video quality they could only observe differences in FL between images captured at the beginning and end of each exercise (Cataneo, 2018). Although this is an unforced error; due to the complexity of collecting dynamic US upon the BF_{LH}, it does impact upon the findings, as the author are then only predicting what occurs during the movement (i.e., between the start and end of each repetition) (Figure 2-17).

Surface EMG has been used by several authors that have sought to characterise the activation patterns of individual hamstring muscles during resistance exercises (Bourne, et al., 2018; Bourne, Williams, et al., 2017; Tsaklis, et al., 2015). The research concluded that between the individual hamstring muscles, there is a degree of selective activation during knee dominant or hip dominant exercises (Bourne, et al., 2018; Bourne, Williams, et al., 2017). During hip dominant exercises e.g., stiff leg deadlift, hip extension, RDL, across both eccentric and concentric actions, the BF_{LH} appears to be preferentially recruited (Bourne, et al., 2018; Bourne, Williams, et al., 2017; Tsaklis, et al., 2015). This is in contrast to knee dominant exercises (e.g., NHE or leg curl), where preferential recruitment of the medial hamstrings (SM & ST) has been reported (Bourne, et al., 2018; Bourne, Williams, et al., 2017; Tsaklis, et al., 2015). However, there is some contention within the literature regarding muscle activation, as contrasting results have also been reported between exercise type and activation profiles (van den Tillaar, et al., 2017; Zebis, et al., 2013). Although this could be explained by the authors not distinguishing between eccentric and concentric actions, which may have interfered with the findings.

With regards to implementation of the NHE within resistance programmes, they are generally prescribed as supra-maximal exercises in a neutral axis – i.e., where the eccentric portion is performed within a controlled descent until a break point is reached. Once athletes are capable to perform the NHE with control, or even able to perform the subsequent concentric phase, the progressive application of load should be applied (Bourne, Duhig, et al., 2017; Duhig, et al., 2019; Pollard, et al., 2019). This progression of load has included the addition of external weight (2.5 – 5 kg), however as the NHE is reliant upon resistance against gravity, an alteration to the performance angle could theoretically increase the force required by the hamstrings to counteract the altered gravitational moment (Figure 6-1). Performing the NHE on an incline or decline surface would essentially manipulate the lever through which the centre of mass, from the knee up is acting, thereby increasing, or decreasing the amount of force required by the hamstrings to control the descent of the centre of mass for any given knee angular displacement. A decline position would result in a greater load at a shorter muscle length, whereas an incline position would reduce the load and potentially involve work at a longer muscle length. Alternative alterations which may have a similar effect at increasing the gravitational moment could include changing arm position, however, this would probably be dependent on an individual's arm length and upper body muscle mass. Additionally, it should also be noted that poor thoracic range of motion could negatively influence the lumbar and pelvis position potentially compromising the performance of the exercise. To date, only a single study has been published, where the authors looked to observe the effect of altering the performance angle of the NHE exercise (Sarabon, et al., 2019). Using inverse dynamics, they identified no differences in peak torque at the knee, between neutral performance 0°, 20° and 40° of inclinations (Sarabon, et al., 2019). Contrastingly, the authors did report significant differences in knee angle occurring at peak knee and hip flexion, with a trend of increasing knee angle with increased performance angle (Sarabon, et al., 2019). Furthermore, normalised EMG of the BF and ST showed a decreasing trend in intensity, with increasing performance angle (Sarabon, et al., 2019). Potentially, highlighting that an incline NHE reduces the intensity of the NHE exercise, however, further research is required to observe the effect of a decline NHE.

Regressions of the NHE typically incorporate the use of elastic bands and other devices, however – the aim of the NHE is to reach a maximum load, to lead to positive adaptations in maximal eccentric strength and BF_{LH} FL. Therefore, utilising elastic bands and other devices to aid performance, will generally result in athletes being able to perform multiple sub-maximal repetitions, which may be suboptimal in improving eccentric strength and BF_{LH} FL. Performing an incline NHE, however, would potentially reduce load applied, allowing athletes to perform the NHE through a greater ROM, while achieving a maximal eccentric stimulus. Therefore, the aims of this study were to investigate the angle at which the NHE is performed and its influence on the kinematic, neuromuscular and the *in-vivo* dynamics of the BF_{LH} throughout the movement. It was hypothesised, from an intensity standpoint, an incline NHE would reduce the involved load or force while a decline NHE would increase the involved load or force. Therefore, it was hypothesised that the kinematic and *in-vivo* mechanics would change to demonstrate this, with either a greater or reduced breakpoint, in addition to differential muscle and fascicle lengthening. Additionally, it was hypothesised the neuromuscular response would support this from an intensity perspective, with greater activation for the decline variation and reduced activation for the incline variation, although neuromuscular patterning between the lateral and medial aspects of the hamstring would be consistent between variations of the NHE.

6.2 Methods

6.2.1 Research design

An observational research design was implemented, whereby participants attended the human performance laboratory on a single occasion for testing. Participants performed variations of the NHE (flat, incline and decline), in a randomised order, while dynamic US images of the self-identified dominant limb's BF_{LH} was concurrently collected. Simultaneous surface EMG of the BF_{LH} and ST was also collected on the contralateral limb only.

6.2.2 Subjects

Thirteen physically active individuals (10 males and 3 females, age 24.7 ± 3.7 years, body mass 79.56 ± 7.89 kg, height 177.40 ± 12.54 cm) with no history of lower-limb injury completed the testing session. All participants reported that they were physically active, with a previous history of performing the NHE within training and were familiar with the exercise. Written informed consent and the results of a health questionnaire (Appendix five), were obtained from all participants prior to testing. The study was approved by the Ethics committee (HSR1819-048) and conformed to the principles of the Declaration of Helsinki (1983).

6.2.3 Procedures

Prior to performing the NHE, participants performed a standardised warm up following the collection of resting US images, which consisted of two sets of ten repetitions of body weight squats, lunges and leg swings. A standardised warm up has been shown to be crucial for the collection of EMG data, with significant differences being identified in RMS amplitude during an isometric action between a warmup and a no-warm up control group (Stewart, et al., 2003).

To perform the NHE, participants were knelt on a specially designed padded Nordic bench (Power lift, Jefferson, IA, USA), with the ankles secured immediately superior to the lateral malleolus by ankle pads which were secured to the bench. From the initial kneeling position, with their ankles secured, arms across the chest and hips extended, participants lowered their body as slowly as possible to a prone position (Bourne, Duhig, et al., 2017). Participants only performed the lowering portion of the exercise, aiming to lower with as much control as possible, until the lowering phase could no longer be controlled and they reached a breakpoint, at which point they were instructed to use their arms to control the decent in a press-up motion. The Nordic bench was positioned across 3 horizontal planes, for NHE performance angles at flat at 0°, incline at 20° and decline at -20° (Figure 6-1). Participants were instructed to perform three repetitions of each variation in a random order, with one-minute rest provided between each repetition and 2-3 minutes between each variation.

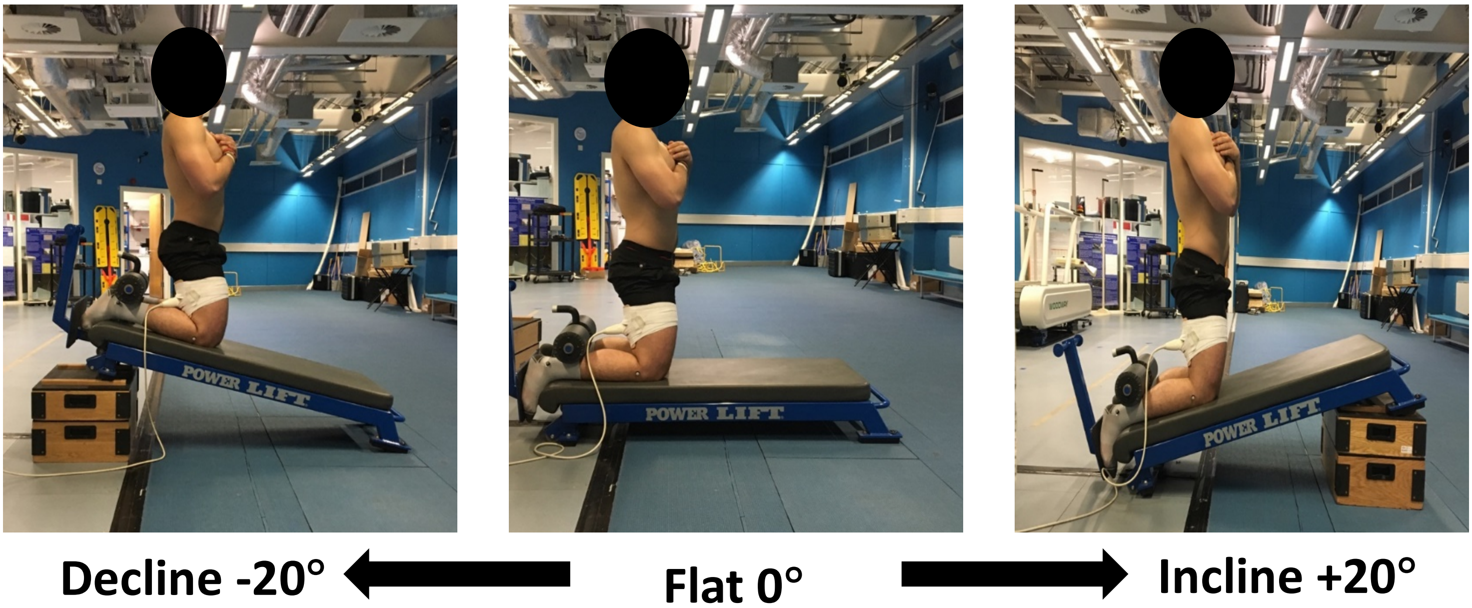


Figure 6-1 Nordic hamstring exercise variations: Flat (0°), Decline (-20°) and Incline (20°).

6.2.3.1 Data collection

6.2.3.2 Resting ultrasound

A full description of how resting US images of the BF_{LH} were collected can be found in Chapter 3 & 4. A measure of resting FL was made, in order to quantify changes in fascicle lengthening during the task relative to FL as suggested previously (Van Hooren, Teratslas, et al., 2020). Participants with shorter resting fascicles, would be expected to demonstrate greater fascicle lengthening, as each sarcomere unit would undergo greater lengthening. Whereas participants with greater resting fascicles, who would in theory have more in-series sarcomeres which means each unit would experience less elongation.

6.2.3.3 Motion capture

To capture 3D lower limb motion data for the NHE variations, infrared Oqus cameras (Oqus 7+, Qualisys AB, Partille, Sweden) were used, operating through Qualisys track manager software (C-motion, version 3.90.21, Gothenburg, Sweden). The NHE bench was surrounded by 10 Oqus cameras mounted in a fixed position in the ceiling, which facilitated standardisation between sessions. Prior to data collection, the Qualisys camera system was calibrated to define the capture volume of testing area. The same calibration process was performed as in the previous chapter (Chapter 5 A & B) and is considered integral to ensure the collection of valid 3D motion data (Chiari, et al., 2005).

Following a successful calibration of the Qualisys camera system, several passive retro-reflective markers (10 mm in diameter), were placed on each of the lateral portions of the participants' legs and trunk. Passive retro-reflective markers were placed upon the medial malleoli, lateral femoral epicondyles, greater trochanter and acromion process. Once the participants were ready for testing, the Qualisys camera system captured three-dimensional motion data for a 15 second period for each NHE trial, sampling at a rate of 250 Hz. Passive retro-reflective marker visibility was checked manually by the tester during the rest periods prescribed between the repetition trials.

6.2.3.4 Electromyography collection

A full description of how EMG data were collected for the BF_{LH} and the ST can be found in Chapter 5 Part A.

6.2.3.5 Dynamic ultrasound

The same linear array probe and US scanner (7.5 MHz, 10 cm, 44 Hz MyLab 70 XVision, Esaote, Genoa, Italy), was utilised to collect dynamic US video clips of the BF_{LH} fascicles during the NHE variations, for the self-identified dominant leg. To define the scanning area, a guideline was drawn upon the participants' skin that corresponded to the BF_{LH} muscle-tendon unit – i.e., line upon the muscle belly between the ischial tuberosity and lateral epicondyle.

A custom designed cast was utilised, that was able to securely house the probe (Figure 62). Double sided adhesive tape and elasticated bandages were used to attach the probe and cast to the posterior thigh. A sufficient amount of conductive gel was applied to the probe prior to being fixed upon the posterior thigh, where it was fixed in orientation to the guideline to achieve an accurate plane to view the BF_{LH} fascicles.

Figure 6-2 Custom designed cast, housing 10 cm ultrasound probe.

Video clips of between 15-17 seconds in length were recorded during each of the NHE variations. Video clips were recorded for a longer duration than the collection of 3D motion data and EMG data. This was because the US videos required commencing prior to starting 3D motion and EMG capture in order for appropriate synchronization to be performed.

6.2.3.6 Synchronization

External synchronization was required to be able to align the dynamic US videos, 3D motion capture and EMG. An external synch pulse was applied, using an additional trigger that connected the US scanner with an open analogue channel which was acquired by Qualisys track manager software. This synch pulse provided a matched time whereby the US images could be synchronised with the 3D motion and EMG data for appropriate analysis and interpretation.

6.2.3.7 Data analysis

6.2.3.8 Motion analysis

Analysis of motion data was performed within Qualisys track manager software, where for specific markers for each repetition, instantaneous hip and knee angles and knee angular velocity were plotted. Raw data was subsequently exported into a custom designed Excel spreadsheet, where movement onset was identified as the moment participants moved > 5° from an average knee angle taken from the first two-seconds of data collection. To identify the moment where participants could no longer control the decent (i.e., breakpoint), a knee angular velocity threshold of 20 deg/s was utilised. Previous literature has utilised a 10 deg/s angular knee extension threshold (Ditroilo, et al., 2013), however, during pilot testing some participants were found to be able to achieve knee angular extension speeds of 10 deg/s with no distinct exponential rise in knee angular velocity. Whereas, when an angular velocity 20 deg/s was achieved there was no return, with a subsequent exponential rise in angular velocity indicating a lack of control, hence a break point (Figure 6-3).

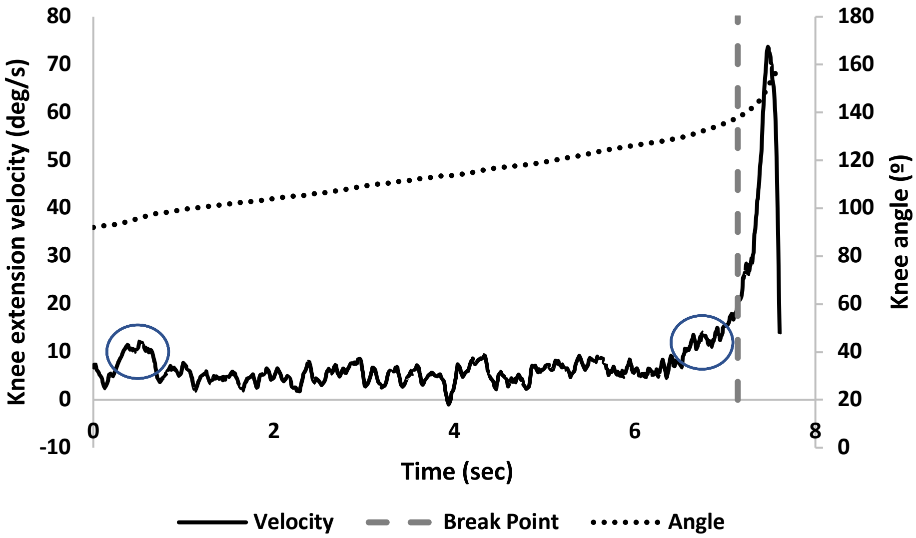


Figure 6-3 Time-Knee extension angle and Time-Knee extension angular velocity graphs, identifying the moment of break point using a 20 deg/s threshold and moments (blue circles) where a threshold of 10 deg/s would have indicated break point.

From the instantaneous hip and knee angles, estimations of BF_{LH} MTU lengths across the NHE repetitions were calculated using regression equations. Hawkins and Hull (1990) identified constant values and algebraic equations that can estimate MTU lengths for all the muscles of the lower limb, including the BF_{LH}. For the BF_{LH}, Hawkins and Hull (1990) identified a quadratic regression equation of;

$$L = C0 + C1\chi + C2\beta + C3\beta^2 + C4\emptyset$$

Equation 6-1 Quadratic regression equation to estimate BFLH FL (Hawkins and Hull, 1990).

Where L represents the normalized muscle length, C0-C4 the constant coefficients and χ, β and Ø the hip, knee and ankle angles respectively. The BF_{LH} constant values are, 1.048, 2.09E-3, -1.60E-3, 0 and 0 for C0, C1, C2, C3 and C4 respectively (Hawkins and Hull, 1990).

6.2.3.9 Electromyography

The raw EMG signals were initially high- and low-pass filtered between 10 and 1000 Hz, as pre-set filters within the Noraxon receiver (Desktop DTS Receiver, Noraxon U.S.A. Inc, Scottsdale AZ, USA), before being exported from the Qualsys track manager software (C-motion, version 3.90.21, Gothenburg, Sweden). This type of initial data processing is performed to remove unwanted artefacts (including noise artefacts (e.g., movement of the cables) and the identification of cardiac signal amplitude). The data was then exported into a custom Excel spreadsheet where further processing and analyses was performed. Within the custom Excel spreadsheet, processing of the EMG data continued with an RMS filter, across a moving average window of 25 ms. This filtering window was chosen as it presented high acceptable reliability for peak and mean EMG values for the BF_{LH} during the GHR (Ripley et al, 2019). Normalization to a perceived maximum value, was not performed within the current study, as comparisons of EMG intensity between variations were made within individuals (i.e., between NHE variations) and not between individuals or groups.

Peak EMG amplitude of both the BF_{LH} and ST were identified across the normalized NHE, in addition, a peak activation ratio of the BF_{LH} to ST was identified. Further measures including iEMG were calculated, from movement onset to the subsequent breakpoint, and EMG RoR – from movement onset to the peak activation amplitude.

6.2.3.10 Resting and dynamic ultrasound

A full description of how resting US images of the BF_{LH} were assessed can be found in Chapters 3 & 4. However briefly, the partial equation [$FL = L + (h \div sin(\beta))$] was utilised as it is a reliable method of estimating BF_{LH} FL and reduces the estimated portion of the fascicle.

Dynamic US videos were analysed using a semi-automated tracking algorithm processed using Ultra track MATLAB graphical user interface (Math-works). Initially, using matched time-points, identified via the applied synchronization pulse, video files were cropped corresponding to the points between movement onset and breakpoint. Following this, in the first US image of each video sequence, a muscle region of interest and fascicle end points were defined. The muscle region of interest was defined as the area between the superficial and deep aponeuroses of the BF_{LH} muscle that was visible in the US image. Muscle BF_{LH} FL was defined as the straight-line distance between the superficial and deep aponeuroses, which was visible in the initial starting position for all participants. This is in contrast to resting

BF_{LH} FL assessments, whereby the whole fascicle was generally not measurable, this is probably due to the change in knee angle.

Changes in BF_{LH} FL were subsequently tracked using an optical flow algorithm with affine optic flow extension (Farris and Lichtwark, 2016), which tracks the transformation between consecutive images in the sequence. Model parameters are computed for a pair of images, where the affine transformation matrix determined new Cartesian coordinates of points defined in the first image to the second image. This process is applied to compute the displacement of the fascicle end points from one image to the next. The Lucas-Kanade optic flow method, has six parameters: V_{XT} – optic flow at the origin (top left corner) in the *x*-direction; V_{yt} – optic flow at the origin (top left corner) in the *y*-direction; *d* – rate of dilation; *r* – rate of rotation; S_1 – shear along the main image axis; S_2 – shear along the diagonal image axis. These parameters can be used to estimate the flow vector (change in position *vx*, *vy*) at specific points in the image (*x, y*) by applying the following first order model:

$$(v_x, v_y) = \begin{bmatrix} x & t & 1 \end{bmatrix} \times \begin{bmatrix} d + s_1 & s_2 + r \\ s_2 - r & d - s_1 \\ v_{xt} & v_{yt} \end{bmatrix}$$

Equation 6-2 Lucas-Kanade optic flow equation.

The algorithm implements a least square fit of the parameters to estimates of the spatial and temporal grey-level gradients on a rectilinear grid within the defined region of interest. To calculate the spatial and temporal gradients, the images first smoothed in both the x and y components by convoluting the image with a one-dimensional (1D) Gaussian mask. To calculate the spatial gradients, the symmetric local difference of the average of the two images is calculated by 2D convolution in the x and y directions using a linear gradient function. Once the parameters for the model are determined, the flow at the individual points can be determined, including, the change in position of any x-y point from one image to the next using the affine transformation. The calculated affine transformation is applied to the Cartesian coordinates of the defined fascicle end points from the first image frame to calculate the new coordinates in the subsequent image sequence, permitting the BF_{LH} FL to be defined for each US image in a sequence.

6.2.4 Statistical analyses

Data are presented as mean ± SD. Normality was assessed by Shapiro-Wilk statistic. Absolute and relative between-trial reliability of peak measures was assessed by CV percentages and a two-way random effects model ICC, with 95% CI determined for both measures of reliability. Minimum acceptable reliability was confirmed using an CV <10% (Hopkins, 2000). A CMC with 95% CI was also performed to analyse the similarity between waveforms by comparing shape, timing and amplitude (Van Hooren, Teratslas, et al., 2020). The ICC and CMC values were interpreted based on the lower bound CI as (<0.50) poor, (0.5-0.74) moderate, (0.75-0.90) good and (>0.90) excellent (Koo and Li, 2016).

Repeated measures analysis of variance (RMANOVA) with Bonferroni post-hoc corrections were conducted to determine if there were significant differences in the kinematic and EMG changes in performance of the NHE between incline, flat and decline. Cohen's *d* ES were also calculated to provide a measure of magnitude of the differences in each variable, interpreted in line with previous recommendations, which defined values of < 0.35, 0.35-0.80, 0.80-1.5 and > 1.5 as trivial, small, moderate, and large respectively (Rhea, 2004). Statistical significance was defined as $p \leq 0.05$ for all tests.

Likely differences between NHE performance angle waveforms for absolute and relative BF_{LH} FL changes were determined by plotting the time normalized average curves for each group along with the corresponding upper and lower 95% CIs to create upper and lower control limits and identifying non-overlapping areas. A secondary method included determining the magnitude of differences between groups by Cohen's *d* ES plotted across 101 nodes. Defined threshold values of values of 0 to 0.35 as trivial, 0.35 to 0.80 as small, 0.80 to 1.5 for moderate and > 1.5 as large differences (Rhea, 2004).

6.3 Results

6.3.1 Kinematics

Descriptive and reliability statistics for both knee angle and MTU length at break point, along with the change from the initial start position to break point are presented in Table 6-1 & 6-2 and Figure 6-4. All kinematic measures and NHE performance angles met acceptable absolute reliability, however, poor-excellent relative reliability was observed for measures of MTU length.

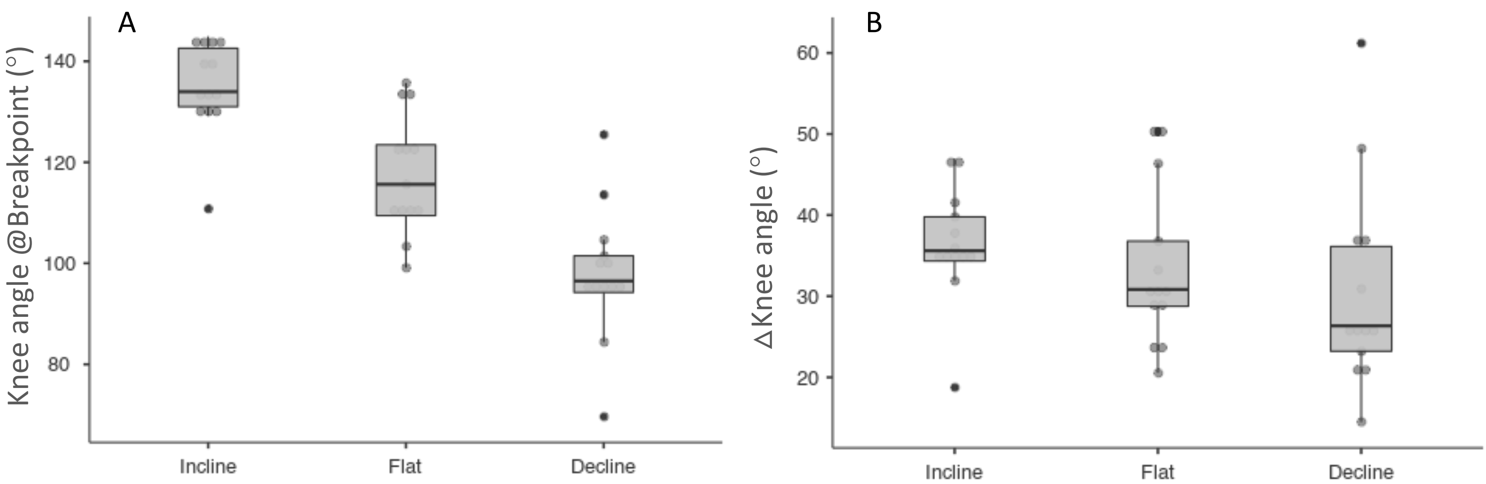
Table 6-1 Descriptive and reliability statistics for kinematic data across NHE variations.

		Mean (SD)	CV%	ICC (95% CI)
Knee angle (°) at Breakpoint	**Incline**	135.08 (9.22)	0.67	0.877 (0.626-0.975)
	Flat	117.67 (11.87)	1.44	0.965 (0.877-0.993)
	Decline	98.04 (13.28)	1.64	0.943 (0.809-0.989)
Change in Knee angle (°)	**Incline**	36.34 (7.02)	2.89	0.787 (0.628-0.955)
	Flat	33.44 (9.87)	1.88	0.908 (0.706-0.982)
	Decline	30.53 (12.65)	1.81	0.951 (0.831-0.991)
Knee angle (°) at Breakpoint relative to the horizontal	**Incline**	64.98 (9.22)	1.39	0.822 (0.725-0.985)
	Flat	61.27 (11.82)	2.80	0.912 (0.857-0.963)
	Decline	61.94 (13.31)	2.60	0.933 (0.849-0.990)
Relative MTU length (%) at Breakpoint	**Incline**	100.95 (2.78)	0.61	0.877 (0.626-0.975)
	Flat	98.64 (2.55)	0.56	0.812 (0.477-0.961)
	Decline	96.77 (3.39)	0.48	0.809 (0.472-0.960)
Change Relative MTU length (%)	**Incline**	7.73 (0.83)	6.45	0.777 (0.408-0.953)
	Flat	7.63 (2.03)	7.78	0.814 (0.482-0.961)
	Decline	7.52 (2.19)	5.29	0.631 (0.176-0.914)
MTU – Muscle-tendon unit				

A statistically significant main effects was observed between NHE performance angles (p <0.001). Post-hoc analysis revealed significant, moderate to large differences were observed for the knee angle at break point across all performance angles (Table 6-2, Figure 6-4), whereas for MTU length at break point there was only a significant, moderate difference observed between incline and decline variants (Figure 6-5). There were no significant differences observed for change in knee angle or change in MTU length.

Table 6-2 Pairwise differences between kinematic measures for each NHE variation.

		Pairwise post hoc p	**Cohen's *d* (95% CI)**	**Effect Size descriptor**
Knee angle (°) at Breakpoint	**Incline Vs Flat**	**0.001**	1.39 (0.14 - 2.59)	Moderate
	Incline Vs Decline	**<0.001**	2.82 (1.20 - 4.38)	Large
	Flat Vs Decline	**0.002**	1.30 (0.06 - 2.49)	Moderate
Change in Knee angle (°)	**Incline Vs Flat**	0.667	0.29 (-0.81 - 1.38)	Trivial
	Incline Vs Decline	0.337	0.51 (-0.61 - 1.61)	Small
	Flat Vs Decline	0.792	0.22 (-0.88 - 1.31)	Trivial
Knee angle (°) at Breakpoint relative to the horizontal	**Incline Vs Flat**	0.801	0.37 (-0.74 – 1.46)	Small
	Incline Vs Decline	0.780	0.42 (-0.69 – 1.51)	Small
	Flat Vs Decline	0.997	0.04 (1.05 – 1.13)	Trivial
Relative MTU length (%) at Breakpoint	**Incline Vs Flat**	0.090	0.70 (-0.44 - 1.81)	Small
	Incline Vs Decline	**0.006**	1.14 (-0.07 - 2.30)	Moderate
	Flat Vs Decline	0.270	0.54 (-0.58 - 1.64)	Small
Change Relative MTU length (%)	**Incline Vs Flat**	0.984	0.06 (-1.03 - 1.15)	Trivial
	Incline Vs Decline	0.945	0.12 (-0.97 - 1.21)	Trivial
	Flat Vs Decline	0.991	0.04 (-1.05 - 1.13)	Trivial
MTU – Muscle-tendon unit				



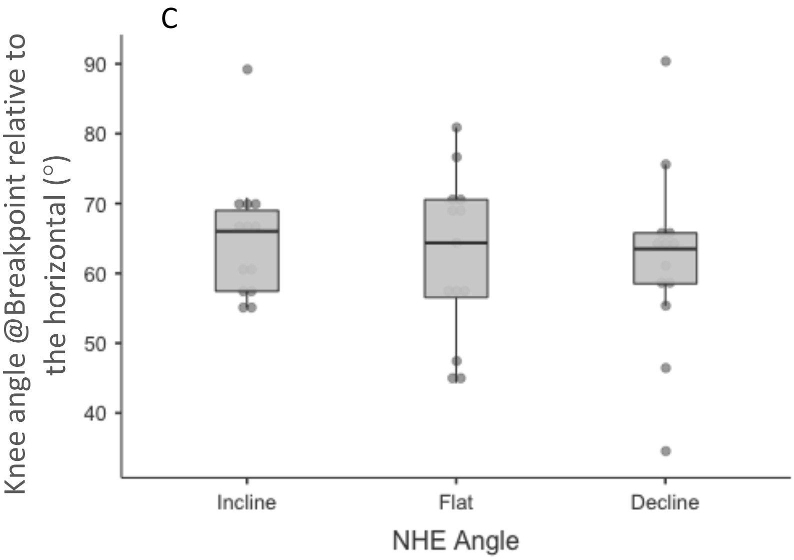


Figure 6-4 Individual, mean, interquartile range, minimum, maximum and outliers within a box and whisker plots for the kinematic measures of knee angle A) Knee angle at break point, B) Change in knee angle, C) Knee angle at break point relative to the horizontal.

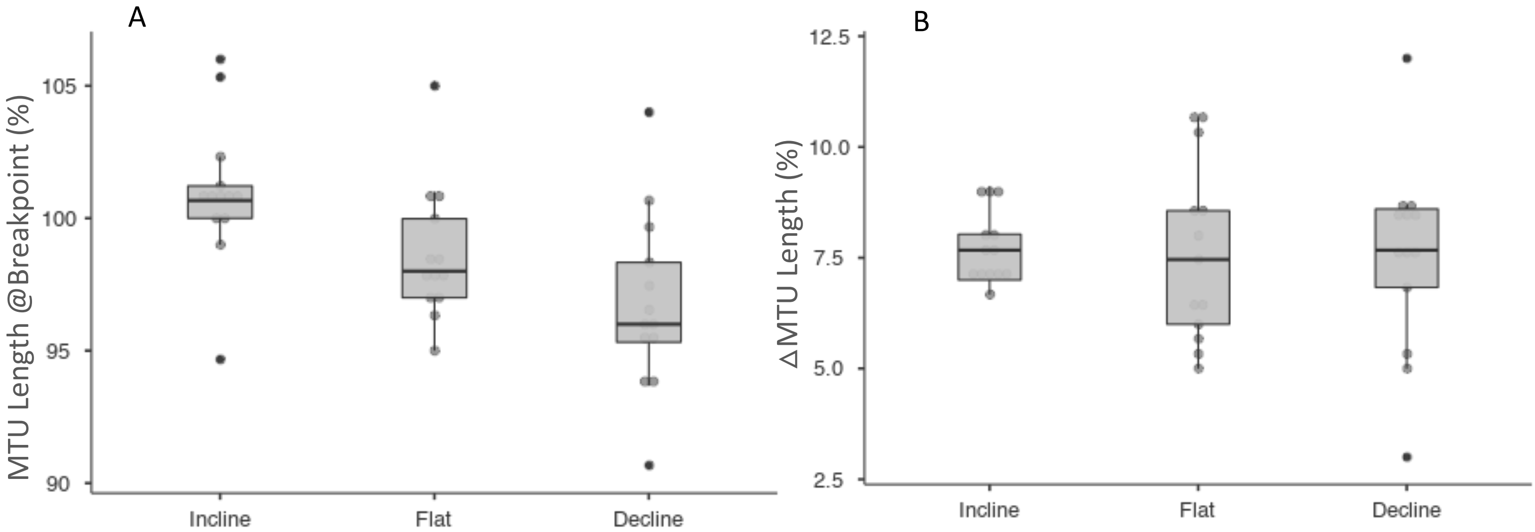


Figure 6-5 Individual, mean, interquartile range, minimum, maximum and outliers within a box and whisker plots for the kinematic measures of Muscle-tendon unit length A) Knee angle at break point, B) Change in knee angle.

6.3.2 Electromyography

Descriptive and reliability statistics for all EMG measures are presented in Table 6-3 & 6-4, and Figure 6-1 & 6-2. The reliability of EMG measures was variable, ROR of EMG for both the BF_{LH} and ST did not meet acceptable absolute reliability with poor relative reliability. All other measures met acceptable levels of absolute reliability, while relative reliability ranged from poor – excellent (Table 6-3).

Table 6-3 Descriptive and reliability statistics for Electromyography data across NHE variations.

		Mean (SD)	CV%	ICC (95% CI)
Peak Bicep femoris EMG (µV)	**Incline**	413.49 (148.75)	1.40	0.957 (0.873 - 0.989)
	Flat	364.70 (124.14)	1.85	0.922 (0.780 - 0.980)
	Decline	324.98 (109.76)	1.17	0.948 (0.849 - 0.987)
Peak Semitendinosus EMG (µV)	**Incline**	352.45 (121.87)	1.21	0.917 (0.769 - 0.979)
	Flat	380.83 (153.72)	1.81	0.908 (0.746 - 0.976)
	Decline	380.27 (153.89)	1.97	0.930 (0.800 - 0.982)
Bicep femoris iEMG (µV/s)	**Incline**	1918.70 (795.78)	9.24	0.715 (0.366 - 0.918)
	Flat	2217.67 (1002.06)	7.73	0.841 (0.594 - 0.958)
	Decline	1801.88 (626.78)	7.80	0.626 (0.235 - 0.887)
Semitendinosus iEMG (µV/s)	**Incline**	1729.51 (675.12)	6.63	0.851 (0.676 - 0.996)
	Flat	2239.70 (1199.53)	4.69	0.975 (0.925 - 0.994)
	Decline	1629.25 (603.70)	6.25	0.860 (0.635 - 0.963)
Bicep femoris EMG ROR (µV/s^{-2})	**Incline**	78.32 (26.30)	15.61	0.638 (0.252 – 0.891)
	Flat	83.54 (33.32)	12.69	0.512 (0.098 – 0.841)
	Decline	84.80 (45.23)	12.10	0.684 (0.282 – 0.901)
Semitendinosus EMG ROR (µV/s^{-2})	**Incline**	72.58 (21.23)	7.94	0.391 (-0.021 - 0.785)
	Flat	77.03 (16.98)	10.33	0.239 (-0.144 - 0.698)
	Decline	81.10 (14.10)	13.97	0.643 (0.259 - 0.893)
Lateral to medial hamstring ratio	**Incline**	1.06 (0.14)	5.78	0.861 (0.637 - 0.963)
	Flat	0.94 (0.06)	2.35	0.955 (0.867 - 0.989)
	Decline	0.79 (0.09)	7.39	0.931 (0.802 - 0.982)
EMG = electromyography, iEMG = integrated EMG & ROR = rate of rise				

A non-statistically significant main effects was observed between NHE performance angles (p =0.651). Post-hoc analysis revealed across all NHE performance angles, there were no significant differences between any single muscle EMG measure with only trivial to small differences observed (Table 6-4, and Figure 6-6). However, significant and moderate to large differences were observed between the lateral to medial hamstring activation ratio (Table 6-4, and Figure 6-7), with an increasing lateral activation, from decline to incline.

Table 6-4 Pairwise differences between electromyographic measures for each NHE variation.

		Pairwise post hoc p	**Cohen's *d* (95% CI)**	**Effect Size descriptor**
Peak Bicep femoris EMG	**Incline Vs Flat**	0.641	0.28 (-0.82 - 1.37)	Trivial
	Incline Vs Decline	0.218	0.53 (-0.59 - 1.63)	Small
	Flat Vs Decline	0.668	0.27 (-0.83 - 1.36)	Trivial
Peak Semitendinosus EMG	**Incline Vs Flat**	0.862	0.17 (-0.92 - 1.26)	Trivial
	Incline Vs Decline	0.867	0.17 (-0.92 - 1.26)	Trivial
	Flat Vs Decline	1.000	0.00 (-1.09 - 1.09)	Trivial
Bicep femoris iEMG	**Incline Vs Flat**	0.681	0.28 (-0.82 - 1.37)	Trivial
	Incline Vs Decline	0.909	0.13 (-0.96 - 1.22)	Trivial
	Flat Vs Decline	0.429	0.38 (-0.73 - 1.47)	Small
Semitendinosus iEMG	**Incline Vs Flat**	0.393	0.47 (-0.64 - 1.56)	Small
	Incline Vs Decline	0.916	0.13 (-0.96 - 1.22)	Trivial
	Flat Vs Decline	0.256	0.48 (-0.64 - 1.57)	Small
Bicep femoris EMG ROR	**Incline Vs Flat**	0.898	0.15 (-0.94 - 1.24)	Trivial
	Incline Vs Decline	0.896	0.16 (-0.93 - 1.25)	Trivial
	Flat Vs Decline	0.996	0.03 (-1.06 - 1.12)	Trivial
Semitendinosus EMG ROR	**Incline Vs Flat**	0.488	0.37 (-0.74 - 1.46)	Small
	Incline Vs Decline	0.734	0.26 (-0.84 - 1.35)	Trivial
	Flat Vs Decline	0.246	0.57 (-0.55 - 1.67)	Small
Lateral to medial hamstring ratio	**Incline Vs Flat**	**0.017**	0.90 (-1.04 - 1.14)	Moderate
	Incline Vs Decline	**<0.001**	1.85 (-0.90 - 1.28)	Large
	Flat Vs Decline	**<0.001**	1.76 (-0.98 - 1.20)	Large
EMG = electromyography, iEMG = integrated EMG & ROR = rate of rise				

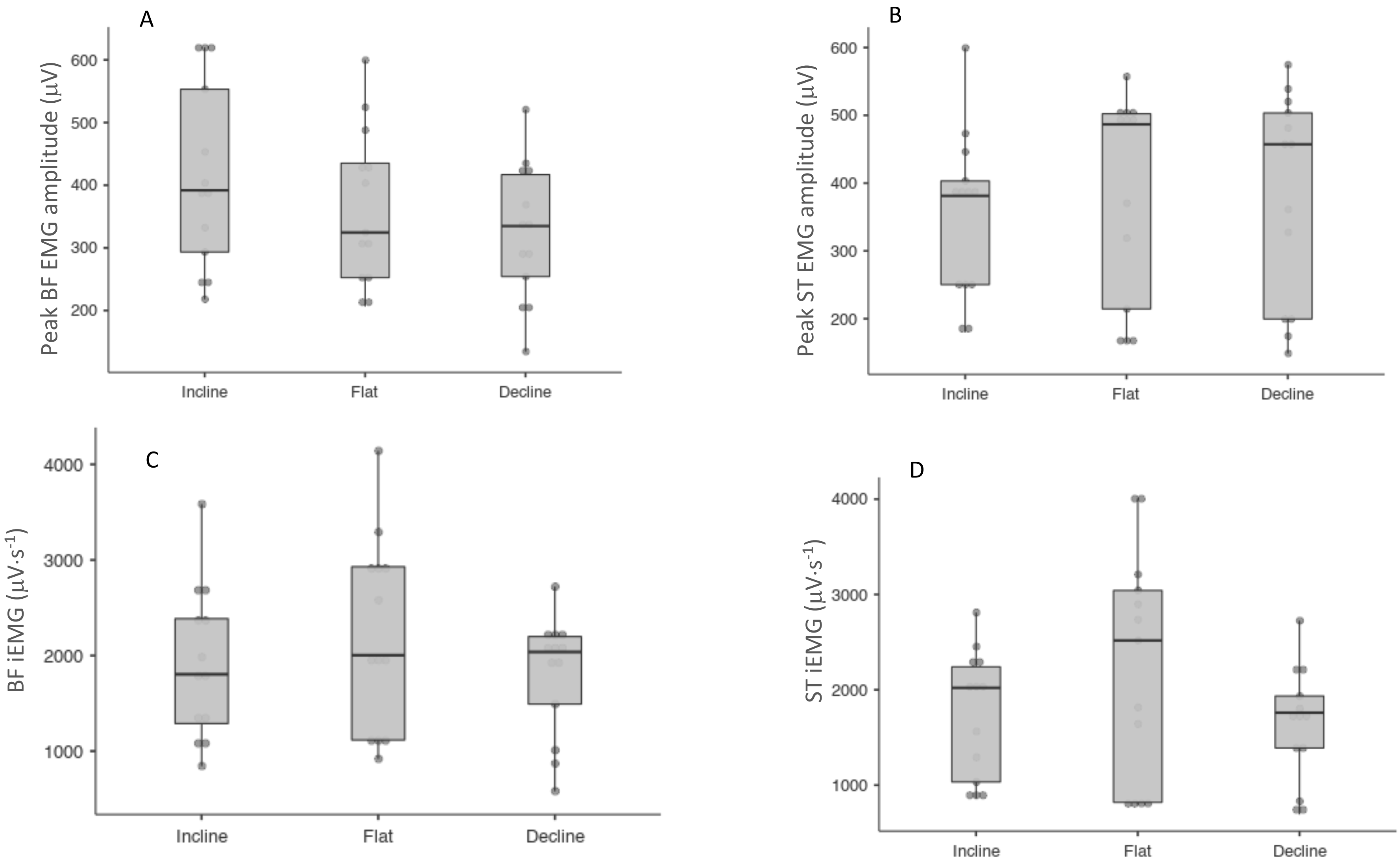


Figure 6-6 Individual, mean, interquartile range, minimum, maximum and outliers within a box and whisker plots for the EMG measures A) Peak BF EMG Amplitude, B) Peak ST EMG amplitude, C) BF iEMG, D) ST iEMG.

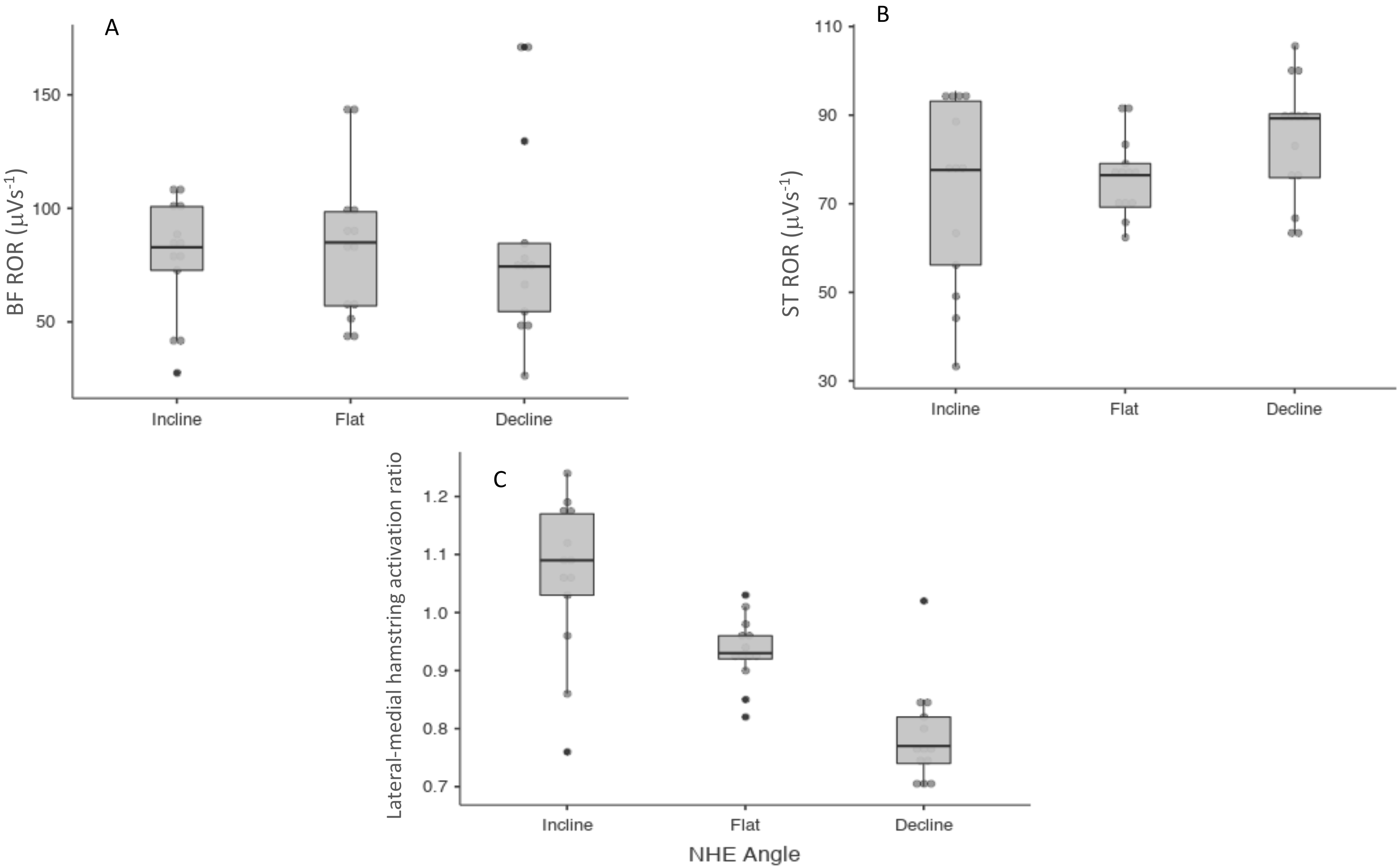


Figure 6-7 Individual, mean, interquartile range, minimum, maximum and outliers within a box and whisker plots for the EMG measures A) Peak BF RoR, B) Peak ST RoR, C) Lateral to medial hamstring activation ratio.

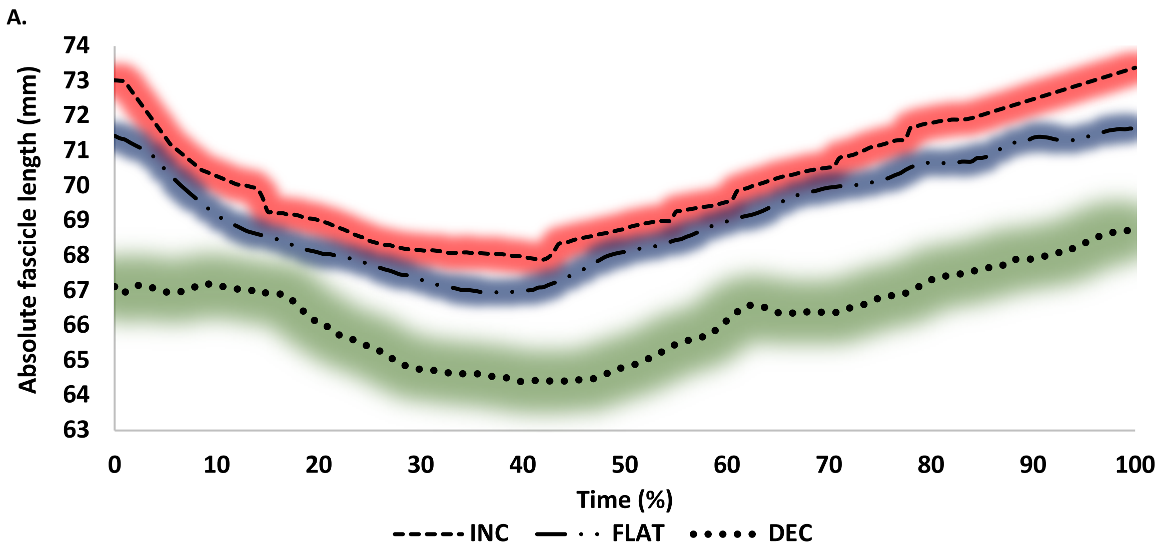
6.3.3 Dynamic ultrasound

As highlighted within the introduction, dynamic US assessment has many difficulties which can influence the observed reliability. However, within the present study moderate-good between trial reliability of the observed across the waveforms was identified (Table 6-5). With meaningful changes from the initial FL at the starting position, to the shortest FL experienced during the NHE to the FL at breakpoint (Table 6-5).

Table 6-5 Absolute and relative between-trial reliability for dynamic ultrasound waveforms and mean and SD absolute FL measurements at the starting position, shortest FL and FL and breakpoint.

	Incline	**Flat**	**Decline**
ICC (95% CI)	0.777 (0.725 - 0.809)	0.817 (0.746 – 0.886)	0.779 (0.733 - 0.825)
CMC (95% CI)	0.769 (0.738 - 0.801)	0.801 (0.742 – 0.840)	0.772 (0.728 - 0.815)
SEM (mm)	0.838	0.890	0.658
Starting FL (mm)	73.02 ± 0.87	71.43 ± 0.92	67.12 ± 1.16
Shortest FL (mm)	67.89 ± 0.92	66.93 ± 0.37	64.40 ± 1.53
FL at breakpoint (mm)	73.38 ± 0.93	71.63 ± 0.59	68.76 ± 1.48

Likely and meaningful differences were observed between NHE performance angle for both absolute and relative FL waveforms (Figure 6-8 & 6-9). For the incline and flat performance angles, absolute FL was likely and to a large magnitude, greater than the decline angle across the entire waveform (Figure 6-8)**,** however, the magnitude of difference between the incline and flat performance angles, was small to moderate, with large differences identified for small portion of the task (Figure 6-8).



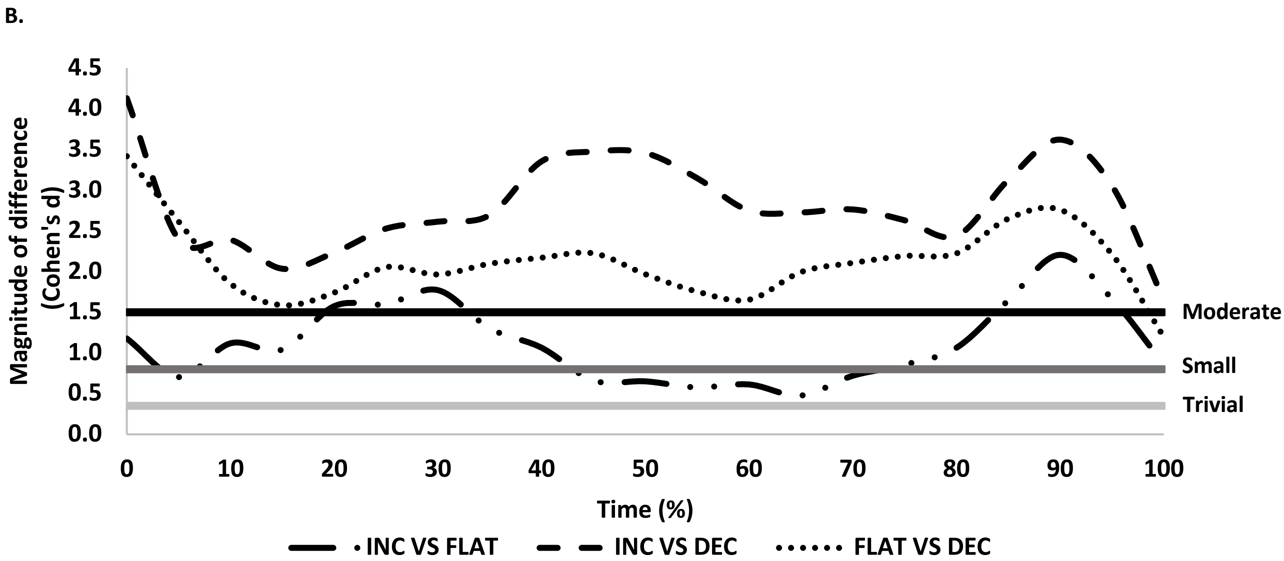
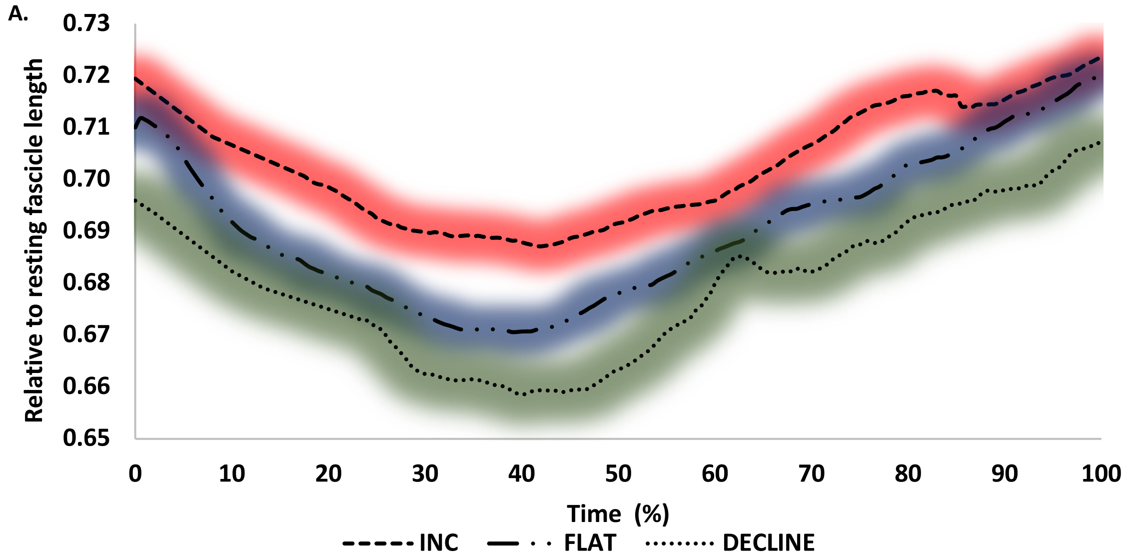


Figure 6-8 A) Dynamic absolute fascicle length changes across performance angles of the NHE, with 95% confidence intervals. B) Magnitude of differences between NHE performance angles.

Where, likely and large meaningful differences were observed across the waveform for absolute FL, when made relative to the individuals resting BF_{LH} FL, the magnitude of differences was smaller – with likely differences only observed across the start-mid range of the waveform for Incline Vs flat and Incline Vs Decline waveforms (Figure 6-9). No likely differences were observed between flat Vs decline, with small to trivial differences identified (Figure 6-9).



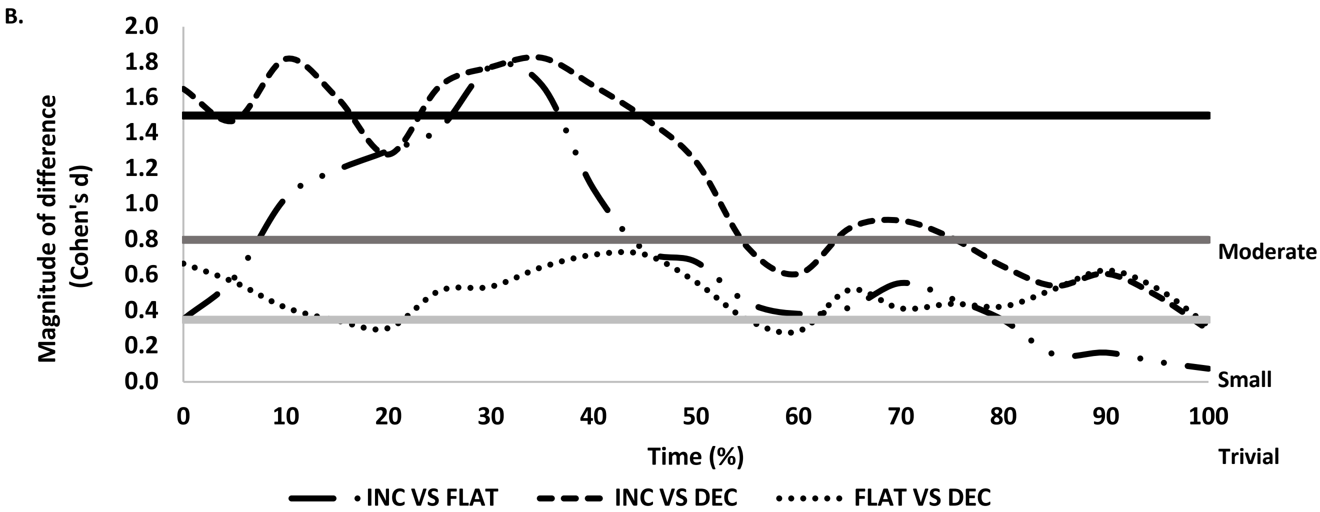


Figure 6-9 A) Dynamic relative fascicle length changes across performance angles of the NHE, with 95% confidence intervals. B) Magnitude of differences between NHE performance angles.

6.4 Discussion

The results of the present study demonstrate that there are differences in the kinematic, neuromuscular and BF_{LH} *in vivo* muscle mechanics between the NHE performance angles, which could indicate there maybe, potential alterations in the adaptive response. As hypothesised and consistent with previous literature (Sarabon, et al., 2019), the incline variation of the NHE resulted in working through a greater ROM – although there is a caveat to this finding. Despite a decreasing trend of a reduced knee angle at break point and MTU length at break point from incline to decline variations, with a significantly greater knee angle at break point and MTU length for the incline variation in comparison to flat and decline variations, the total movement performed between variations is similar, with only trivial to small, non-significant differences identified between the change in knee angle and MTU length. Additionally, using break point angle relative to the horizontal, as an indicator of gravitational moment, there was also trivial to small, non-significant differences highlighting that the forces experienced may be similar between variation. However, by altering the starting position (i.e., NHE performance angle), utilising the incline NHE variation within a training programme, could lead to a potentially larger positive adaptive response to the BF_{LH} architecture and eccentric strength as at the moment of eccentric overload (i.e., breakpoint), the hamstring complex would be working at a greater muscle length. The increased muscle length could be the only reason as to why it may be superiorly effective, as it could be suggested that as a breakpoint was still achieved in each variation, that the gravitational moment could be identical between variations.

Training at a long muscle length resulted in a greater positive adaptation in BF_{LH} FL, whereas in contrast, training at a short muscle length resulted in a greater positive adaptation for eccentric hamstring strength (Guex, Degache, et al., 2016). Although, the duration of the training intervention was only three weeks, thus the adaptation observed after week three could have been lower than any eventual adaptive response with a longer training duration, or a longer delay between finishing the intervention and post-testing. Franchi et al. (2017) highlighted the potential that a more pronounced mechanical stretch could be applied to single sarcomeres or fascicles during large ROM tasks, influencing serial sarcomere distribution and eventual architectural adaptations (Blazevich, Cannavan, Coleman, & Horne, 2007). This may signify that an incline NHE, could lead to larger positive adaptations in BF_{LH} FL in contrast to the traditional flat NHE or decline variations. Although a similar effect may not be seen with eccentric hamstring strength, where a traditional or even decline NHE may provide a greater positive effect.

A recent systematic review on the effect of ROM on muscle development found that for lower body musculature working with a greater ROM results in similar or greater increases in muscle size or volume (Schoenfeld and Grgic, 2020). This potentially indicates that training at a greater end ROM, as found within the incline NHE variation, could lead to an increase in hamstring muscle volume. Moreover, when performing the NHE at longer muscle lengths there is a rise in early torque production (Hegyi et al., 2019), however, if an eccentric break point is achieved during the NHE, then peak torques attained will likely to be similar (Sarabon, et al., 2019). Although the same trend may not be seen in the present study as this early rise torque production was produced by changing the hip angle (90° hip flexion). Furthermore, the peak torques achieved between both the NHE performed at both 0° and 90° of hip flexion, were similar with non-significant, trivial differences identified (A Hegyi, Johan Lahti, et al., 2019), although this finding is not consistent within the literature as Sarabon, et al. (2019), found significant, moderate to large differences in peak hip and knee joint torques, when simultaneously altering the knee and hip NHE performance angles. Therefore, it could be presumed that the adaptations in eccentric hamstring strength could be similar between the NHE performance angles, as the peak eccentric torques, or gravitational moments are similar. When implementing the NHE within athlete populations' programmes, the aim should be increasing eccentric hamstring strength (Bourne, et al., 2015; Bourne, et al., 2018; Shield and Bourne, 2018; Timmins, Bourne, et al., 2016). However, an increase in hamstring muscle volume could be a desired response for certain populations, including weaker individuals with small muscle volume, youth athletes and for aesthetic athletes. Although, within the present study it is not ROM that has increased with the incline variation, it is in fact the muscle length and BF_{LH} FL that have increased.

The EMG response demonstrated that between NHE performance variations, there were minimal differences between individual muscles response (peak amplitude, iEMG and RoR) (Figure 6-6 & 6-6). However, there were significant, moderate-large differences observed for the lateral-medial hamstring activation ratio, with the incline variation demonstrating a greater lateral hamstring emphasis, moving to a medial emphasis for both the flat and decline variations. Typically, the NHE if thought of as being a medial dominant exercise with regards to activation ratios (Bourne, et al., 2018; Bourne, Williams, et al., 2017), this holds true within the present study, as both the flat and decline variations presented greater ST (i.e., medial dominant). Contrastingly, it was observed that the incline NHE variation was more lateral dominant exercise, which could indicate that the incline variation could be more effective training tool than the traditional flat NHE. One possible explanation as to why the incline NHE variation is more lateral dominant exercise could be the change muscle length, with muscle length being one of the key determinants of the magnitude of muscle activation (A. Vigotsky, Halperin, Lehman, Trajano, & Vieira, 2017). However, the results of the present study, contrast that of previous research, where significant, small-moderate decreases in relative activation of the BF and ST were observed with increasing slope (i.e. the greater the incline, the lower the relative activation) (Sarabon, et al., 2019). Additionally, those results also showed minimal between muscle difference between with an equal activation between both the medial and lateral components of the hamstrings (Sarabon, et al., 2019). However, these could be explained by methodological differences, for example as the current study did not to normalize EMG amplitudes, as with previous studies. The rationale behind the present study not normalizing EMG amplitudes was that as comparisons of EMG intensity between variations were made within individuals and not between and any normalization procedure could have influenced the observed result (Burden, 2010). Therefore, a more accurate reflection of the task intensity and patterning could be provided from non-normalized EMG amplitudes. Interestingly, the largest individual variation of lateral-medial hamstring activation ratios was found within incline variation, with the lowest finding identical to that of the mean activation ratio of the decline variation. The observed individual variation could be the participants' NHE ability, i.e., even during the incline variation, high intensities were achieved as indicated by reaching a break point very early within the movement and at a short muscle length for some individuals. Additionally, the individual variations in EMG could in fact be the result of individual preferential coordination strategies performed with the task and between the tasks (Avrillon, et al., 2018; Avrillon, et al., 2020), with the potential for variations in force-sharing strategies within and between the muscles of the hamstrings.

Motor control has been identified as a potential risk factor of HSI incidence (Pizzari, et al., 2020), although the current level of evidence for motor control being an influencing factor on HSI incidence is limited. It would be presumed that due to the requirement of high rate of force or torque development during the terminal phase of sprinting (Chumanov, et al., 2011; Thelen, Chumanov, Best, et al., 2005; Thelen, Chumanov, Hoerth, et al., 2005), an elevated neural drive or neuromuscular response, i.e. high RoR of EMG, would be essential and training of such qualities could be a factor in exercise selection. The present study highlights that across all variations of the NHE the RoR of EMG was similar, indicating all variations could improve upon the rapid neuromuscular functioning of the hamstrings, along with rate of torque development. Although to date, the RoR of EMG within the hamstrings has never been observed in resistance exercises, an increase in rate of torque development via resistance training could be related to elevations in RoR in early phases of onset of muscle action (Aagaard, et al., 2002). Although prospectively, eccentric rate of torque development and onset of EMG activation, has been shown to have a minimal influence on the risk of HSI incidence (van Dyk, Bahr, et al., 2018).

To the authors knowledge, this is the first study to effectively assess hamstring muscle architecture dynamically across an entire exercise movement. It has been highlighted, that there are several difficulties when attempting to dynamically image hamstring muscle architecture including: field of view, spatiotemporal resolution, transducer design and image quality. Despite these difficulties, the current study found moderate levels of between trial reliability using both ICCs and CMCs (Table 6-5). This highlights that both the image and the imaging quality were of a high enough standard (i.e., described methods permitted appropriate imaging, despite identified difficulties). The moderate level of reliability reported also highlights the ability of the semi-automated tracking system (Ultra track) to dynamically assess the BF_{LH} during exercise – it is the first time this system has been used to assess the hamstrings.

The results of the dynamic US imaging highlighted novel findings surrounding the eccentric actions of the hamstring musculature. Across all the NHE variations for both absolute and relative FLs, approximately 40% of the ROM involved fascicle shortening (i.e., concentric action), after which there was fascicle lengthening (i.e., eccentric action) (Figure 6-7 & 6-8). Between the three NHE variations across the entire ROM, the incline variation involved a greater absolute and relative FL in comparison to the other variations, with likely and large differences across the waveform. In contrast, the decline NHE variation involved a lower absolute and relative FL across the entire ROM. One explanation for the difference in FLs, could be from the initial starting positions – as the incline NHE commences at a more extended knee angle and a greater MTU length it is unsurprising that the FL would be greater in comparison to the other variations. The greater degree of fascicle lengthening observed within the incline NHE, could indicate that the incline NHE maybe preferential when attempting to achieve architectural adaptations of the BF_{LH}. Specifically, when observing absolute FL at the end range of the NHE (~80% time), the incline NHE FL was likely and meaningfully greater than both the flat and decline NHE variations. Although, the same trend was not represented when observing relative FL, with no likely differences at the same end range of the NHE (i.e., small and trivial in magnitude). Therefore, the potential of each NHE variation to provide a stronger positive training effect may not be presumed.

Similar to the previous work that dynamically imaged the hamstrings during the NHE, the present study found that from the start and end of the movement the BF_{LH} fascicles went through a lengthening process, regardless of NHE performance angle (Cataneo, 2018). Although, between the studies it is difficult to compare the magnitude of change, it does support previous work observing that the application of load to a muscle has minimal influence upon the magnitude of fascicle lengthening – especially when observing the changes relative to a resting FL as within the present study. As the greatest change in FL across all variations was occurring at the mid-end ranges of motion (>50% range), as the majority of the lengthening within the early stages is taken up by the elastic components (Ando et al., 2016; Ando, et al., 2018). However, the present study only observed the dynamic changes occurring within the BF_{LH}, while it has been identified previously; that within the quadriceps muscle group, dynamic FL changes during eccentric exercise are individual muscle specific, typically as a result of different muscle compositions (i.e., percentage of muscle to tendinous tissue) (Ando, et al., 2016; Ando, et al., 2018). Therefore, it could be presumed that within the hamstring complex during eccentric exercise, such as the NHE, there is variety of dynamic FL changes occurring, potentially highlighting the need for a multi-factorial training process.

The current study is not without limitations, firstly, the present data only includes two-dimensional images, which fails to capture the complex interactions between transverse and longitudinal muscle strains and the potential rotation around the longitudinal axis that can occur under voluntary actions (Van Hooren, Teratslas, et al., 2020), potentially questioning the validity that two-dimensional imaging provides. However, three-dimensional practices provide further challenges for gathering data, especially for the present study as the University of Salford has only one 10-cm probe, hence effective three-dimensional data collection would not have been possible. Additionally, the present study utilised "Ultratrack" software for automated FL tracking, with previously established reliability and validity (Farris and Lichtwark, 2016). It does not, however, account for the potential for fascicle curvature. Further, it was not possible to report PA, as the software provides the angle between the identified fascicle and the horizontal axis, which was not always in line with the deep aponeurosis, leading to inaccuracies. Within dynamic US research, "Ultratrack" is the most popular tool within the literature, which is suggested to be attributed to it being freely available, while having a well-designed user interface (Van Hooren, Teratslas, et al., 2020). This is despite new or other established approaches to computational tracking approaches having improved reliability and validity, although it has been suggested that developers should take a similar approach in making freely available code, within a useable interface (Van Hooren, Teratslas, et al., 2020). It is also important to note that the sample is not exceptionally large (n = 13), this in-turn negatively effects the studies power demonstrating that any ES d <0.84 is meaningfully underpowered (<0.80) (Jamovi project (2018) Computer Software, retrieved from https://www.jamovi.org). It was the original plan to boost this number once collection of the training intervention (Chapter 6) was complete, however, due to the circumstances surrounding Co-vid19 it became impossible/impractical to collect this data at the time.

Despite the present study being successful at imaging the BF_{LH} during exercise, overcoming the numerous difficulties, the design and function of the medial hamstrings (SM and ST), could make it nearly impossible to effectively image each of the hamstring muscles dynamically during exercise. However, as technology improves further research should look to observe the dynamic fascicle changes that occur across all of muscles within the hamstring complex. In addition, it would be prudent to assess dynamic fascicle changes in hamstrings across a wider range of exercises incorporating multiple modes of action – of high interest would be running, as this may present details on their functioning, which has been an ever-present academic argument (Maniar, Schache, Heiderscheit, & Opar, 2020). Additionally, future dynamic US studies should look to group participant by either eccentric strength or NHE ability (e.g., Nordbord), which was not assessed within the present study. This information could highlight differences in technique or strategy of loading, specifically within the NHE – despite the present study highlighting that the application of load to a muscle has minimal influence upon the magnitude of fascicle lengthening, an individual's level of strength could alter the velocity and strategy of loading.

6.5 Conclusions

The present study found that between NHE variations, there were significant and meaningful effects upon kinematic, neuromuscular and BF_{LH} *in vivo* muscle mechanics. Changes to the NHE performance angle manipulates the lever arm through which the centre of mass, from the knee up is acting, thereby increasing, or decreasing the amount of force required by the hamstrings. Although, the observed changes in instantaneous knee angle and MTU length at break point could in fact be related to the altered starting position of each performance angle, with the start position of the incline NHE having the most extended knee angle, while the decline position has the lowest knee angle. The observed change in starting position, which influenced knee angle and MTU length could also explain the observed differences in the neuromuscular contributions of the BF_{LH} and the ST to the task, as the increased working muscle length found within the incline NHE could be in fact due to altering neuromuscular contributions of the hamstrings to be BF_{LH}, which contrasts the flat and decline NHE, as well as previous literature (Bourne, et al., 2018; Bourne, Williams, et al., 2017).

Additionally, the differences in BF_{LH} *in vivo* muscle mechanics, when compared to a resting FL, which could be an indicator of eccentric hamstring strength, found likely, meaningful differences within the early-mid range of movement (0-40% time), where greater fascicle shortening was observed within the decline and flat NHE variations. This observation could be explained by the decline and flat NHE variations requiring the contractile components within the BF_{LH}, to take up more slack which would be present within the elastic component (i.e., distal tendon), which is under less strain within the early stages of the movement. Despite these differences early within the movement, there were no likely or meaningful differences identified between variations at the mid-end range (40-100% time). Therefore, all variations result in a similar magnitude of relative fascicle lengthening, which may indicate that similar positive adaptations in eccentric hamstring strength and BF_{LH} FL would be attained from there utilisation.

6.6 Linking paragraph

Across training interventions, all modalities appear to be effective at increasing eccentric hamstring strength, with eccentrics (sprint and NHE) and long length traditional exercises being superior at specifically increasing BF_{LH} FL. Although it should be highlighted that despite these modalities being effective at increasing eccentric hamstring strength and BF_{LH} FL and reducing the incidence of HSIs, we can never totally prevent HSI occurrence and continue to identify methods to minimise the risk of HSIs. However, the interventions performed to increase eccentric hamstring strength and BF_{LH} FL that have been performed to date are not without their limitations, such as extremely high volumes, inappropriate intensity, or lack ecological validity as the investigators are looking at the effect of single exercise. However, within Chapter 6 it appeared to potentially favour an incline NHE, with regards to kinematic, neuromuscular, and *in vivo* muscle mechanics, for adaptations to eccentric hamstring strength and BF_{LH} FL. Although, the differences were not conclusive, as when dynamic FL was made relative to resting FL, the magnitude of change in relative BF_{LH} FL during the NHE variations was similar (incline, decline and flat).

Although practice has been implementing sprint training for some time, it is only recently made its way into research supporting its utilisation (Freeman, et al., 2019; Mendiguchia, et al., 2020). However, consistent to previous studies observing hamstring resistance training (i.e., NHE), these were again not without their limitations. One of the main limitations found across all studies, is that they typically lack ecological validity and a controlled, consistent, resistance training programme is not reported or defined. Therefore, it would be prudent to observe the effect of a controlled resistance training programme which incorporates a long length, hip dominant, traditional exercise, with additional sprint or NHE training – as part of a multi-modal approach as used within elite practice.

7 Study 5 – Effect of additional Nordic hamstring exercise or sprint training on the modifiable risk factors of hamstring strain injuries and performance.

7.1 Introduction

Across the literature, training interventions that have attempted to reduce HSI incidence, have aimed to mitigate the influence of the modifiable risk factors of HSI (i.e. eccentric hamstring strength and BF_{LH} FL), by targeted exercises, such as the NHE (Bourne, Duhig, et al., 2017; Ribeiro-Alvares, et al., 2018; van der Horst, et al., 2015; van Dyk, et al., 2019) or as a combination of exercises (i.e. FIFA 11/11+ warm up protocol (Thorborg, et al., 2017)). Incorporating the NHE has a meaningful ability to decrease the occurrence of HSI, however, the effectiveness of any intervention modality relies upon the compliance of the athletic population (Bourne, et al., 2018), with ≥75% compliance showing superior effectiveness within the literature (chapter 2-5). Low levels of compliance within studies that have utilised the NHE as part of training interventions have frequently been reported due to the effect of DOMs and/or poor athlete support. This is despite only a moderate level of DOMs being reported within NHE training interventions (Bourne, Duhig, et al., 2017; Cuthbert, et al., 2019). Furthermore, the NHE; one of the most extensively researched eccentric hamstring exercises, is continually poorly adopted within elite European soccer (Bahr, et al., 2015), despite showing superior effectiveness (chapter 2-5). Bahr, Thorborg and Ekstrand (2015), cited high levels of both player and coach complaints when implementing the NHE. One possible explanation is that many players and coaches do not fully understand the potential benefits of implementing the NHE, with many unconvinced of key intervention outcomes (i.e., the NHE reduces injuries, increases player availability, return to play sooner post-HSI) (Bahr, et al., 2015).

Currently, the NHE has been a key focus of training research by observing its effect on one or more of the modifiable risk factors of HSI (i.e., eccentric strength, muscle architecture) (Bourne, Duhig, et al., 2017; Bourne, et al., 2018; Duhig, et al., 2019). Interventions that have utilised the NHE have shown large and significant positive adaptations in both eccentric strength capabilities (isokinetic and Norbord) and BF_{LH} muscle architecture (i.e., increased BF_{LH} FL and decreased pennation angle) (chapter 2-6, 2-7). A recent systematic review and meta-analysis highlighted that the application of the NHE has generally coincided with extremely high volumes, with many interventions progressing to ≥100 repetitions per week - prescribing sets of between 8-12 repetitions (Cuthbert, et al., 2019). This is despite the NHE being classified a 'supra-maximal' eccentric exercise, of a greater intensity than an equivalent concentric action. Furthermore, as the aim of including the NHE should be to increase the force generating potential of the hamstrings (i.e., increase strength), the current prescription would not fall within the repetition and volume guidelines for the implementation of strength training (Sheppard and Triplett, 2016). More recent research has adopted a low volume approach to NHE training (2 x 4 repetitions performed twice per week (Presland, et al., 2018)), increasing eccentric hamstring strength and BF_{LH} FL, to a similar magnitude as higher volume equivalents, while being more aligned with volume recommendations for strength training.

As a result of the continued low compliance of NHE training, a natural progression of practice and research is to investigate the possibility of training that could be more agreeable or available for both athletes and coaches. One example could be sprint training, as it has been hypothesised there could be a similar imposed demand of fascicle lengthening (i.e. eccentric muscle action), while coinciding with the maximal activation patterns during the swing phase (Chumanov, et al., 2011; Higashihara, et al., 2015, 2016, 2018; Higashihara, et al., 2019; Thelen, Chumanov, Best, et al., 2005), which is potentially indicative of the desired adaptive response (i.e. increased eccentric strength and BF_{LH} FL). Furthermore, maximal sprinting has the potential to strengthen the elastic properties of connective tissue, increase motor unit activation, increase passive tension of the muscle-tendon complex and improve cross bridge mechanics, which are all associated with the occurrence of injuries and overall athletic performance (Haugen, Seiler, Sandbakk, & Tønnessen, 2019).

To date, two studies have observed the effects of a sprint-based training on the modifiable risk factors for HSI (Freeman, et al., 2019; Mendiguchia, et al., 2020). Freeman and colleagues (2019) observed a positive adaptive response in eccentric hamstring strength from sprint training. Both sprint and NHE training provided a small but significant, positive response to eccentric hamstring strength – although on closer inspection, the NHE training group, who started stronger, displayed a greater adaptive response than the weaker sprint group (9.8- Vs 6.2%Δ) (Freeman, et al., 2019). This indicates that although both groups improved, the NHE was superior (Freeman, et al., 2019). Although, it should be noted that this study was performed across a short duration of four-weeks, where there was no control of other resistance training – both of which could influence the observed response. More recently, Mendiguchia, et al. (2020) performed a similar study by observing the effect of either the NHE or sprint training upon BF_{LH} architecture. Interestingly, the sprint training group had a moderate, positive increase in BF_{LH} FL, whereas the NHE training only resulted in a small, positive increase in BF_{LH} FL (Mendiguchia, et al., 2020), with a 16.21 vs. a 7.38% change, respectively. Although there could be some methodological explanations to these findings. Firstly, the NHE training could be described as being sub-optimal, as there was no progression of eccentric intensity, following a previously established protocol (first six weeks of the study by Peterson et al. (2011)). Secondly, the sprint training intervention was quite intensive with high volumes, even in comparison to the earlier study by Freeman (2019), although even attempting to equate volumes would be close to impossible with a number of complex variables that would need to be considered (including, muscle action type, muscle action time under tension, stride length, stride frequency, repetitions and distances).

Improvements in athletic performance (e.g., strength, sprinting and jumping) are also a key if not the primary consideration when programming for athletes. It is well documented that sprint-based training can improve athletic tasks (Haugen, et al., 2019; Markovic, Jukic, Milanovic, & Metikos, 2007; Rumpf, Lockie, Cronin, & Jalivand, 2016). Likewise, improvements in both sprint and jump performance have also been observed following a NHE intervention (Askling, et al., 2003a; Clark, Bryant, Culgan, & Hartley, 2005; Ishoi, et al., 2018; Krommes, et al., 2017), although the research is inconclusive regarding athletic performance improvements (Suarez-Arrones, et al., 2019). It has been hypothesized that increases in athletic performance, as a result of NHE interventions, are the result of an increased force generating capacity during hip extension (Morin, et al., 2015), although it is not a well-established theory (Suarez-Arrones, et al., 2019). Therefore, both sprint and NHE training modalities have the potential to increase performance in athletic tasks, as well as mitigating the risk of HSIs via the improvement of the modifiable risk factors. However, some researchers continually neglect the fact that the aim of the NHE is to mitigate the risk of HSIs, via improvement in both eccentric hamstring strength and BF_{LH} FL. Thus, conducting a randomized, parallel training study where additional sprint or NHE training is implemented, with measures of hamstring strength, architecture, and performance in dynamic tasks (i.e., sprint, strength and jump performance) taken before and after, would be insightful for practitioners with respect to identifying potential best practice and how multiple elements could compliment a complete training programme.

7.1.1.1 Aims and hypothesis

The aim of the present study was to determine the effect of a short-term (seven-week) intervention with supplemental sprint or NHE, imbedded within an ecologically valid training programme (control training vs control**+**NHE, control**+**sprint), on the magnitude of adaptations to the modifiable risk factors, i.e., BF_{LH} muscle architecture and eccentric hamstring strength. In addition, a further aim was to observe the effect of the training intervention on the nature of adaptations to overall athletic performance (sprint, CMJ and lower body strength).

It was hypothesised that using a multi-modal approach, with the supplemental NHE or sprint training, would provide the greatest adaptive response to both modifiable risk factors of HSI (BF_{LH} muscle architecture and eccentric hamstring strength), postulating the greatest adaptive response attained from the NHE training. In addition, it was hypothesised that for CMJ performance the NHE training group would improve upon the countermovement phase, due to an increase in eccentric hamstring capabilities, whilst all groups would improve both absolute CMJ measures (e.g., jump height, take-off velocity), in addition to measures made during the propulsive phase (e.g., propulsion force and impulse). It is further hypothesised that for sprint-based measures, the sprint training group would have the greatest adaptations in performance in comparison to other training groups. Finally, it is hypothesised that there would be no difference in lower body strength, as all groups would be following the same control resistance training programme (not including the NHE).

7.2 Methods

An intervention study design was employed for the present study (Figure 7-1), pre-intervention testing was completed for all participants, with a group returning on a second occasion to determine between-session reliability. All subjects were initially randomly allocated into training groups and then completed a comparable 7-week period of resistance training. One group performed the resistance training as a control, without the addition of the NHE or sprints, while the remaining groups performed an identical resistance training programme with the addition of sprint or the NHE.

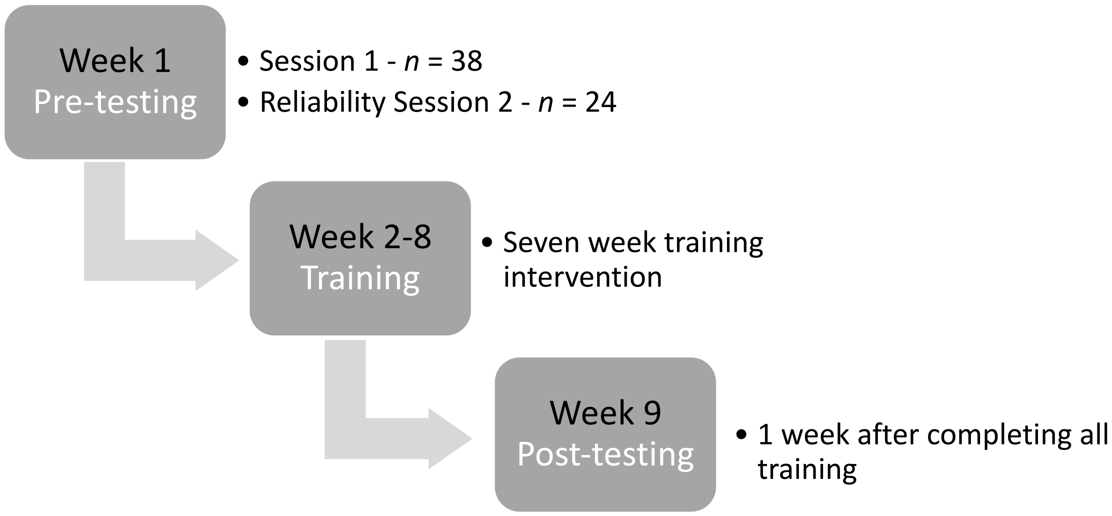


Figure 7-1 Schematic diagram of pre-testing, seven-week intervention and post-testing.

7.2.1 Subjects

38 collegiate athletes who participated in regular team sports (football, futsal, rugby union, rugby league, ice hockey, American football, basketball, netball), from university (collegiate) to semi-professional level sports participation. Participants playing season varied between either pre- or in-season. Participants were randomly allocated to the three training groups using a random number generator; **Nordic** *n* = 15 (7 female, 8 male), age = 21.40 ± 2.64 years, height = 1.74 ± 0.04 m, mass 76.95 ± 14.20 kg, **Sprint** *n* = 13 (4 female, 9 male), age = 22.15 ± 2.54 years, height = 1.74 ± 0.05 m, mass 70.55 ± 7.84 kg, **Control** *n* = 10 (2 female, 8 male), age = 23.50 ± 2.95 years, height = 1.75 ± 0.09 m, mass 77.66 ± 11.82 kg. All participants were of good overall health based on the completion of a Health Questionnaire (Appendix five). All participants reported having a history of resistance-based training, including the NHE, however not regularly (less than once/week) applied within the previous 6 months. Furthermore, all subjects reported having between 1-2 years of sprint or running based technical coaching which had been delivered during sport-based training. The study was approved by the institutional ethics committee (HSR1819-103), and all participants had both read a Participant Information Sheet and provided written informed consent prior to testing. The study also conformed to the principles of the Declaration of Helsinki (1983).

Based on investigating changes in both BF_{LH} architecture and eccentric hamstring strength, G*Power (version 3.1.9.2) was used *a-priori* to calculate sample size, please observe the power and sample size statistics below. (Faul, Erdfelder, Lang, & Buchner, 2007). An effect size of 1.2 was utilised as this magnitude of change was close to what was used within previous literature (Pollard, et al., 2019), additionally, as it was approximately the mean value of 50% of previously reported effect sizes (chapter 2-6 & 2-7) (Bourne, Duhig, et al., 2017; Duhig, et al., 2019; Pollard, et al., 2019; Presland, et al., 2018). These effect sizes were chosen, as the methods employed were similar to those used within the present study.
Minimum acceptable Power – 0.80
α – 0.05
Effect size – 1.2
***a-priori* sample size** – 12 /group

7.2.2 Procedures

7.2.2.1 Training Programme

Participants in the control and both the intervention groups completed an identical lower limb resistance training programme, performed twice per week. Each resistance training session consisted of three lower limb exercises, where the training volume remained constant across the training intervention, whilst intensity was manipulated (Table 7-1). Immediately post-training, using a numeric scale of 1-10, a rating of perceived exertion (RPE) was obtained from all participants. Approximately 24-hours post-training, using a numeric pain scale of 1-10, a score for DOMS was attained for all participants.

Table 7-1 Lower limb resistance training programe, including sets x reps and estimated one repetition maximum percentages, performed by the control and intervention groups across the seven-week training intervention.

Day 1							
Weeks	**1**	**2**	**3**	**4**	**5**	**6**	**7**
Power clean	3 x 3	3 x 3	3 x 3	3 x 3	3 x 3	3 x 3	3 x 3
	80%	85%	90%	75%	80%	85%	90%
Back Squat	3 x 3	3 x 3	3 x 3	3 x 3	3 x 3	3 x 3	3 x 3
	80%	82.50%	85%	75%	80%	82.50%	85%
Reverse lunge	3 x 6	3 x 6	3 x 6	3 x 6	3 x 6	3 x 6	3 x 6
	70%	72.5%	75%	70%	72.5%	75%	77.5%
Day 2							
Weeks	**1**	**2**	**3**	**4**	**5**	**6**	**7**
Mid-thigh pulls	3 x 3	3 x 3	3 x 3	3 x 3	3 x 3	3 x 3	3 x 3
	80%	85%	90%	75%	80%	85%	90%
Romanian deadlift	3 x 6	3 x 6	3 x 6	3 x 6	3 x 6	3 x 6	3 x 6
	70%	72.5%	75%	70%	72.5%	75%	77.5%
Reverse lunge	3 x 6	3 x 6	3 x 6	3 x 6	3 x 6	3 x 6	3 x 6
	70%	72.5%	75%	70%	72.5%	75%	77.5%

In conjunction to the control resistance training programme, the intervention groups were prescribed either additional sprint or NHE training at the start or end of each training session (Table 7-2), respectively. The NHE volume was maintained across the seven-week intervention, in accordance with the low volume recommendations by Presland et al. (2018). Where participants were observed to have sufficient strength to completely control the movement in the final 10-20° of knee extension during the NHE, they were then required to hold a weight plate to ensure supramaximal exercise intensity was maintained (2.5 kg increments) (Bourne, Duhig, et al., 2017; Duhig, et al., 2019). It should be noted that the findings of the previous chapter were not applied within this study firstly due to the time of submission of the NSCA grant and analysis of the data from the previous chapter. Furthermore, due to the limited literature observing the effect of sprint-based training and the use of a multi-modal approach to training it was decided that the NHE stimulus should only be progressed as per the previous literature. The sprint training group initially experienced incremental increases in sprint volume for the first four-weeks, following which the volume was maintained. The rationale behind this initial increase in volume, is that if participants were not familiar with regular maximal sprint training, a gradual increase in training volume will limit the large spikes in training load in order to reduce the risk of HSI occurrence (Malone, et al., 2018). Sprint training was split across the week, where one training day commenced from a static three-point stance whereas on the second training day participants utilised a rolling start, aiming to accelerate into the sprint similar to the prescription by Freeman et al. (2019). A certified strength and conditioning coach was present at all training sessions, providing verbal feedback on the participants' performance and technique throughout the intervention.

Table 7-2 Additional training performed by the NHE or sprint intervention groups across the seven-week training intervention, including sets x reps.

Day 1 & 2							
Weeks	**1**	**2**	**3**	**4**	**5**	**6**	**7**
NHE	2 x 4	2x 4	2 x 4	2 x 4	2 x 4	2 x 4	2 x 4
Sprint	4 x 25 m	5 x 25 m	6 x 25 m	7 x 25 m	7 x 25 m	7 x 25 m	7 x 25 m

The study aimed to control for any other resistance training performed by the subjects, advising that outside the prescribed programme no further lower-limb resistance training could be performed. Only an individual's sport-specific and upper body resistance training was permitted.

7.2.2.2 Data collection

All testing commenced with resting US imaging of the BF_{LH}. For the collection of BF_{LH} muscle architecture two US devices were utilised, two separate devices were required as some participants were assessed away from the University of Salford biomechanics lab. Therefore, similar to previous sections, a 10 cm linear array with a depth resolution of 67 mm (7.5 MHz, MyLab 70 XVision, Esaote, Genoa, Italy) was used when testing was performed at the biomechanics lab. However, as some testing was performed off site, a secondary more portable device was utilised (Echo Blaster, Telemed UAB, Vilnius, Lithuania). As the findings of Chapter 4 showed that the FOV is crucial in the assessment of BF_{LH} muscle architecture, two images (6-cm + 6-cm) were taken longitudinally across the mid-belly of the muscle making sure they followed a linear path and were congruent to one another. This method has been used previously to assess BF_{LH} FL and shown to be highly reliable (Brennan, et al., 2017; Pimenta, et al., 2018). Utilising the shorter 6-cm probe in this way permitted an extended FOV of 12-cm, which allowed for direct measurement of all fascicles for all off site assessments.

Following muscle architecture assessment, participants performed a standardised dynamic warm-up consisting of body weight squats, forward and reverse lunges, submaximal squat jumps and CMJs. Three maximal effort CMJs, with a one-minute rest between trials was assessed using a Kistler force platform, sampling at 1000 Hz, with data collected via Bioware 5.11 software (type 9286AA, Kistler Instruments Inc. Amherst, NY, USA). Subjects were instructed to stand still for the initial one second of data collection (McMahon, Murphy, Rej, & Comfort, 2017; McMahon, Suchomel, Lake, & Comfort, 2018) to enable the subsequent determination of body weight (vertical force averaged over one second). Raw unfiltered, force-time data was exported for subsequent analysis. For the CMJ, subjects were instructed to perform the jumps as fast and as high as possible, whilst keeping their arms akimbo. Any jumps that were inadvertently performed with the inclusion of arm swing or leg tucking during the flight phase were omitted and additional jumps were performed after one minute of rest.

The assessment of eccentric knee flexor strength was performed using the Nordbord device (Vald Performance, Newstead, Australia), which has been used in the literature previously (Bourne, Duhig, et al., 2017; Bourne, et al., 2015; Franchi, Ellenberger, et al., 2019; Freeman, et al., 2019; Opar, Piatkowski, et al., 2013; Opar, et al., 2015; Timmins, Bourne, et al., 2016). Within the present study, participants knelt upon a padded board, with ankles secured superior to the lateral malleolus by two individual ankle braces. Attached to the ankle braces were uniaxial load cells (50 Hz), allowing for the force generated by the knee flexors during the NHE to be measured. Participants were instructed to perform one set of three maximal NHE repetitions. The instructions to participants were to gradually lean forward at the slowest possible speed while maximally resisting the movement with both limbs, keeping the trunk and hips in a neutral position with the hands held across the chest. Strong verbal encouragement was provided for each subject in order to provide a maximal effort. An acceptable trial required the force output to reach a distinct peak (indicative of maximal eccentric strength), followed by a rapid decline in force, when the participant was no longer able to resist the gravitational forces (Bourne, Duhig, et al., 2017; Bourne, et al., 2015; Franchi, Ellenberger, et al., 2019; Freeman, et al., 2019; Opar, Piatkowski, et al., 2013; Opar, et al., 2015; Timmins, Bourne, et al., 2016).

For the isometric mid-thigh pull (IMTP), the procedures and guidelines previously described were used (Comfort et al., 2018). Each subject adopted a posture that they would use for the start of the second pull phase of the clean, resulting in knee and hip angles of $139.2 \pm 2.8^\circ$ and $149.9 \pm 3.2^\circ$, respectively. All subjects were familiar with this position, through previous performance of weightlifting exercises within training. Joint angles were measured using hand-held goniometer and recorded for standardization. A steel bar which was identical to an Olympic lifting bar, was in a fixed position above the force platform (type 9286AA, Kistler Instruments Inc. Amherst, NY, USA), at a height which replicated the start of the second pull phase of the clean. Subjects stood on the force platform with their hands fixed to the bar with lifting straps (Comfort, et al., 2018). Two warm-up trials were performed with one-minute rest provided, at 50% and 75% of the participants perceived maximum effort. Once participants had adopted an appropriate position, a countdown of “3,2,1, Pull!” was provided. Minimal pretension (<50 N) was permitted, to ensure minimal slack, prior to initiation of the pull, participants were instructed to pull against the bar, as hard and as fast as possible, pushing their feet into the ground (Comfort, et al., 2018). Two maximal effort trials were performed for approximately five seconds, with strong verbal encouragement provided. Between trials, peak force was required to be within 250 N of each other.

Prior to completing the sprint assessment, two 20 m practice sprints at 50- and 75% of perceived maximum intensity, which also served as a brief familiarisation period. Three maximum effort trials of the 20 m sprint were performed, with brief rest periods of two minutes prescribed between trials. Instructions were provided to participants to initiate the sprint from a stationary two-point, split start and to perform a maximal effort throughout the full 20 m (Yeadon, Kato, & Kerwin, 1999). Any sprint trials that were initiated with a countermovement were discarded and supplementary sprint trials were recorded. Brower single-photocell electronic timing gates (Draper, Utah, USA) were placed at 0 m, 10 m and 20 m increments along an indoor running track or 3G AstroTurf, with each emitter and reflector spaced 2 m apart at approximately hip height (Yeadon, et al., 1999). Although the initial pair of timing gates were placed at 0 m, the participants started 0.3 m behind this point (Yeadon, et al., 1999). Sprint times for each distance were recorded via a handheld computer and the successful maximal effort sprint trials for each participant were taken forward.

7.2.2.3 Data Analysis

A full description of how resting US images of the BF_{LH} were assessed can be found in Chapter 3. However, briefly Equation 7-1 was utilised for fascicle that were not within view of the single 10-cm image, as it is a reliable method of estimating BF_{LH} FL and reduces the estimated portion of the fascicle. Relative FL was also established, by dividing the absolute FL measure by the participants femur length, the same method that was utilised within Chapter 5A & B.

$$FL = L + (h \div sin\,(\beta))$$

Equation 7-1 Fascicle length estimation partial measure equation.

Where L is the observable fascicle length, h is the perpendicular distance between the superficial aponeurosis and the fascicles visible end point and β is the angle between the fascicle and the superficial aponeurosis.

Raw force-time data for the CMJ, IMTP and NHE was analysed in Microsoft Excel (Excel 2016, Microsoft, Washington, USA). For the CMJ, velocity of centre of mass at take-off was determined as a measure of performance (take-off velocity) (GL Moir, 2008, 2014), take-off velocity was used in place of jump height, as its measurement error is typically lower. Take-off velocity was determined by dividing vertical force data (minus body weight) by body mass and then integrating the product using the trapezoid rule. The onset of movement for each CMJ trial was considered to have occurred 30 milliseconds prior to the instant when vertical force had decreased by five times the SD of body weight, as derived during the one second silent period (McMahon, Murphy, et al., 2017; McMahon, Rej, & Comfort, 2017; McMahon, et al., 2018). CMJ take-off was identified when vertical force decreased below five times the standard deviation of the force during the flight phase (residual force) (McMahon, Murphy, et al., 2017; McMahon, Rej, et al., 2017; McMahon, et al., 2018). The CMJ phases were identified using the previously established methods (McMahon, Murphy, et al., 2017; McMahon, Rej, et al., 2017; McMahon, et al., 2018). Briefly, the unweighting phase of the CMJ was considered to have occurred between the onset of movement and the instant of peak negative centre of mass velocity. The braking phase of the CMJ was defined as occurring between the instant of peak negative centre of mass velocity and zero centre of mass velocity. The propulsion phase of the CMJ was deemed to have occurred between the instant centre of mass velocity exceeded 0.01 $m{\cdot}s^{-1}$ and the instant of take-off. Braking peak force was defined as the maximum value attained during the braking phase. Propulsion mean force was determined as the mean force during the propulsion phase, while impulse was calculated as the area under the net force-time curve (minus body weight) for the propulsion phase using the trapezoid rule (McMahon, Murphy, et al., 2017; McMahon, Rej, et al., 2017; McMahon, et al., 2018). Countermovement displacement and time, was calculated by the combined time or displacement of centre of mass from the initial standing quiet period to the instant of zero centre of mass velocity, achieved at the end of the braking phase. Therefore, including the combined time and displacement of centre of mass during the unweighting and braking phases.

For the IMTP, peak absolute and relative net force was determined as the maximum forces recorded from the whole force-time curve during the IMTP trials (Comfort, et al., 2018).
For the NHE, consistent with the IMTP, peak force was determined as the maximum forces recorded from the whole force-time curve. Movement onset was determined as the point when the force increased above a 5 N absolute threshold, whereas the movement was finished when the vertical force decreased below a 5 N absolute threshold. Total and active impulse were determined by integrating the whole force-time curve and the active portion of the force-time curve (movement onset-finish), respectively. Mean force was determined as the average force across the active portion of the force-time curve. Time to peak force was determined as the time between movement onset and peak force, while repetition time was determined as the time between movement onset and movement finish.

The mean performance of the trials for each assessment was used for further analysis.

7.2.3 Statistical Analyses

7.2.3.1 Reliability and measurement error

All data was first tested using the Shapiro-Wilk test to check if it satisfied parametric assumptions. A two-way random-effects model ICC and CV with corresponding 95% CI, was used to determine the absolute. The ICC values were interpreted based on the upper and lower bound CI as (<0.50) poor, (0.5-0.74) moderate, (0.75-0.90) good and (>0.90) excellent (Koo and Li, 2016).

A subsample performed two PRE-testing sessions (n=20), to determine the between-session reliability of each variable. As parametric assumptions were met, a repeated measure analysis of variance (RMANOVA), with post-hoc pairwise comparisons with Bonferroni correction was performed to determine if there was a learning effects between trials (within each session) and between testing sessions (within each week of testing).

The SEM and SDD for each variable were calculated to establish random error scores between two testing sessions performed within week 1 of the intervention (Figure 7-1). The SEM was calculated using the following formula [3], where SD_{pooled} represents the pooled SD across the two testing sessions:

$$SD_{pooled} \times \sqrt{1-ICC} \quad [3]$$

The SDD was calculated using the following established formula [4]:

$$\left(1.96 \times \sqrt{2}\right) \times SEM \quad [4]$$

As test-retest reliability and measurement error was established for all variables, any observed changes in performance that exceed the associated measurement error would likely be 'true' changes induced by completing the training programme.

7.2.3.2 Pre to Post changes

Data obtained at pre was taken forward to perform comparisons at post training, as parametric assumptions were met for all measures, between-session (i.e., pre and post) differences in the modifiable risk factors (BF_{LH} FL and eccentric hamstring strength), jump and IMTP measures were determined via a RMANOVA with post-hoc pairwise comparisons with Bonferroni correction applied. Hedge's *g* ES was calculated to provide a measure of the magnitude of the differences in each variable between trials, sessions and groups and interpreted in line with previous recommendations which defined values of < 0.35, 0.35-0.80, 0.80-1.5 and > 1.5 as trivial, small, moderate, and large, respectively (Rhea, 2004).

Unfortunately, due to unforeseen circumstances, sprint testing was not able to be performed upon the control group. Therefore, for sprint testing between-session (i.e., pre and post) differences in sprint times were determined via a RMANOVA with post-hoc pairwise comparisons with Bonferroni correction applied. Hedge's *g* ES was calculated to provide a measure of the magnitude of the differences in each variable between trials.

All statistical analyses performed using SPSS software (version 25; SPSS Inc. Chicago, IL, USA) with the alpha level set at $P \leq 0.05$. All other statistical analyses will be conducted in Microsoft Excel.

7.3 Results

7.3.1 Reliability and measurement error

Chapter 3 found that all measures of muscle architecture (FL, PA and MT) were reliable when using the 10-cm FOV, with acceptable absolute and relative reliability. Similarly, between session reliability for the present study found that using both the 10-cm FOV and 12-cm (6- + 6-cm) FOV methods were also reliable with comparable absolute and relative reliability (7-3).

Table 7-3 Descriptive and reliability statistics for BF_{LH} FL for both 10- and 12cm FOV.

	Mean	SD	CV%	ICC (95% CI)	SEM	SEM%	SDD	SDD%
10-cm FOV – Absolute FL (cm)	9.80	0.16	1.65	0.980 (0.938 - 0.995)	0.17	1.73	0.47	4.80
10-cm FOV – Relative FL	0.22	0.01	3.22	0.975 (0.929 - 0.989)	0.00	1.42	0.01	3.94
12-cm FOV - Absolute FL (cm)	9.55	0.15	1.57	0.927 (0.864 - 0.979)	0.21	2.21	0.58	6.12
12-cm FOV - Relative FL	0.23	0.02	6.88	0.911 (0.844 - 0.959)	0.01	3.26	0.02	8.63

The present study found that absolute and relative peak eccentric hamstring force (N) possessed good levels of relative reliability (Table 7-4).

Table 7-4 Descriptive and reliability statistics for eccentric hamstring strength using the Nordbord.

	Mean	SD	CV%	ICC (95% CI)	SEM	SEM%	SDD	SDD%
Peak Force (N)	326.71	16.18	4.95	0.953 (0.886-0.981)	12.51	3.83	34.67	10.61
Relative peak Force (N/kg)	4.23	0.08	1.89	0.932 (0.877-0.987)	0.19	4.61	0.54	12.77

Across the measures of athletic performance, five of the six CMJ performance measures were found to possess good-excellent relative reliability, and acceptable absolute variability (Table 7-5), with countermovement displacement displacing poor-good relative reliability.

Table 7-5 Descriptive and reliability statistics for measures of athletic performance of the countermovement jump.

	Mean	SD	CV%	ICC	SEM	SEM%	SDD	SDD%
Countermovement Time (s)	0.44	0.01	3.31	0.926 (0.823-0.970)	0.02	4.03	0.05	11.17
Peak Braking Force (N)	1962.74	27.20	1.39	0.912 (0.792-0.964)	109.82	5.60	304.41	15.51
Countermovement Displacement (cm)	0.23	0.02	8.93	0.644 (0.293 - 0.842)	0.09	36.82	0.24	102.07
Mean propulsion Force (N)	1589.25	12.64	0.80	0.981 (0.952-0.992)	39.42	2.48	109.28	6.88
Mean propulsion impulse (Ns)	192.74	2.15	1.12	0.991 (0.976-0.996)	3.07	1.59	8.50	4.41
Take off velocity (m/s)	2.51	0.03	1.23	0.973 (0.933-0.989)	0.04	1.56	0.11	4.34

Both measures attained on the IMTP were found to possess excellent relative reliability and achieved acceptable absolute variability (Table 7-6). With reasonable SEM and SDD values, meaning both measures could be used to assess change in IMTP peak force.

Table 7-6 Descriptive and reliability statistics for measures of athletic performance of the isometric mid-thigh pull.

	Mean	SD	CV%	ICC	SEM	SEM%	SDD	SDD%
Peak Net Force (N)	1743.15	6.46	0.37	0.976 (0.932 - 0.991)	127.64	7.32	212.28	12.18
Peak Relative Net Force (N/kg)	24.40	0.07	0.27	0.966 (0.928 - 0.990)	2.25	9.20	3.73	15.30

Sprint measures of 0-10 m, 0-20 m and 10-20 m split times were found to achieve good to excellent relative reliability and acceptable absolute variability (Table 7-7).

Table 7-7 Descriptive and reliability statistics for measures of athletic performance of sprints.

	Mean	SD	CV%	ICC	SEM	SEM%	SDD	SDD%
0-10 m (s)	1.97	0.01	0.34	0.959 (0.899 - 0.983)	0.02	0.97	0.05	2.70
0-20 m (s)	3.22	0.01	0.20	0.980 (0.949 - 0.992)	0.02	0.62	0.06	1.72
10-20 m (s)	1.26	0.01	0.71	0.897 (0.759 - 0.958)	0.02	1.55	0.06	4.28

7.3.2 Pre- to Post-intervention changes.

7.3.2.1 Fascicle length

For absolute and relative BF_{LH} FL, a non-significant time×training interaction was observed (p = 0.236). Pairwise comparisons revealed significant ($p < 0.001$) and moderate-large increases in FL for all training groups (Figure 7-1 & 7-2). The Nordic and sprint training groups displayed the large increases in both absolute and relative FL (Table 7-8).

Table 7-8 Pairwise comparisons of Bicep femoris fascicle length for all training groups.

Absolute bicep femoris long head fascicle length (cm)					
	Pre	**Post**	**Mean Difference (%)**	**Hedge's *g* (95% CI)**	***p***
Nordic	9.85 ± 1.20	11.12 ±	1.26 (12.83)	0.89 (-0.19 – 1.94)	<0.001
Sprint	9.76 ± 0.74	10.71 ± 0.85	0.94 (9.67)	0.92 (-0.25 – 2.05)	<0.001
Control	9.66 ± 0.93	10.54 ± 0.94	0.88 (9.09)	0.70(-0.60 – 1.96)	<0.001
Relative bicep femoris long head fascicle length					
	PRE	**POST**	**Mean Difference (%)**	**Hedge's *g* (95% CI)**	***p***
Nordic	0.22 ± 0.02	0.25 ± 0.02	0.03 (12.78)	0.94 (-0.15 – 2.00)	<0.001
Sprint	0.24 ± 0.02	0.26 ± 0.02	0.02 (9.24)	0.84 (-0.32 – 1.96)	<0.001
Control	0.21 ± 0.03	0.23 ± 0.03	0.02 (9.23)	0.48 (-0.80 – 1.73)	<0.001

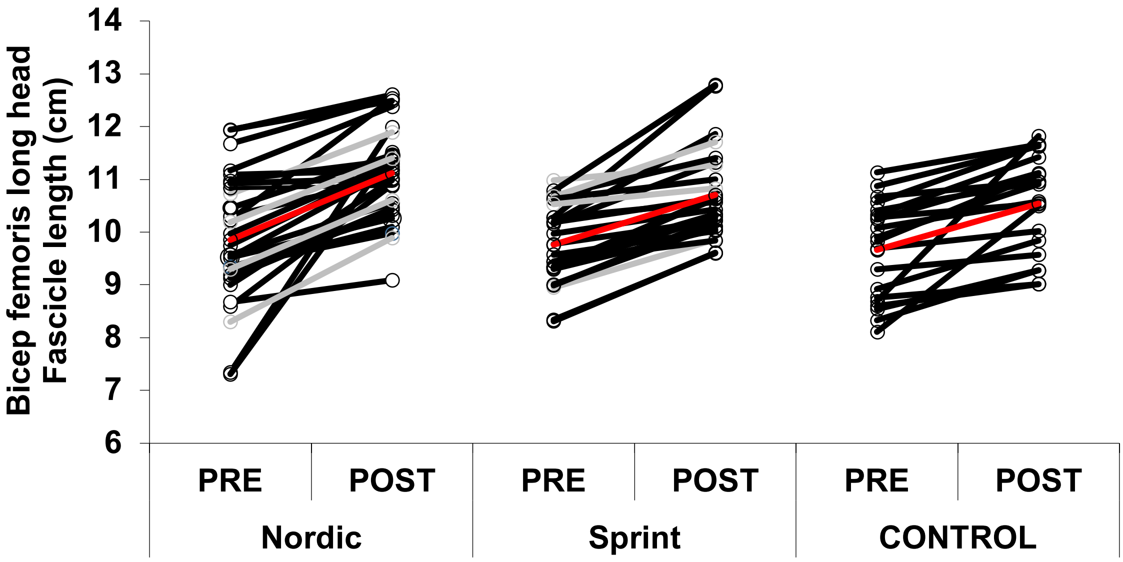


Figure 7-2 Pre- and Post-intervention individual (black) and mean (red) changes for absolute bicep femoris fascicle length. Grey plots signify 12-cm FOV.

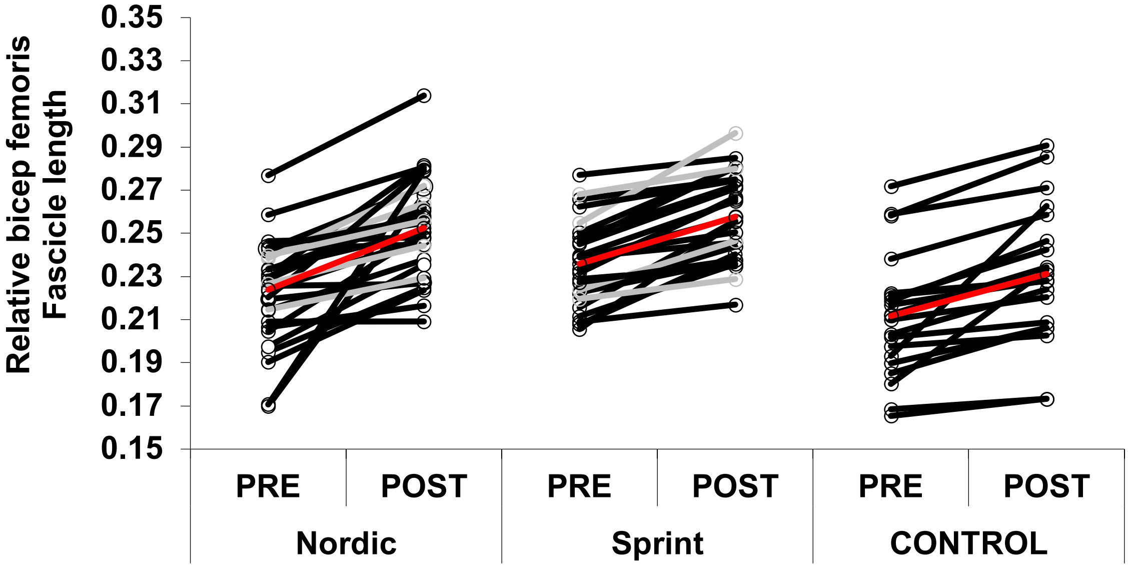


Figure 7-3 Pre- and Post-intervention individual (black) and mean (red) changes for relative bicep femoris fascicle length. Grey plots signify 12-cm FOV.

7.3.2.2 Eccentric hamstring strength

Peak and relative peak force demonstrated a significant time×training interaction ($p < 0.01$). Pairwise comparisons revealed significant ($p < 0.001$) and moderate-large increases in absolute and relative peak force for all training groups (Figure 7-3 & 7-4). The Nordic and sprint training groups displayed large increases in both absolute and relative peak force (Table 7-9).

Table 7-9 Pairwise comparisons of eccentric hamstring measures for all training groups.

Absolute peak eccentric hamstring strength (N)					
	Pre	**Post**	**Mean Difference (%)**	**Hedge's *g* (95% CI)**	***p***
Nordic	317.71 ± 61.93	431.28 ± 59.86	113.58 (35.75)	1.40 (0.24 – 2.52)	<0.001
Sprint	295.80 ± 72.90	386.00 ± 54.51	90.20 (30.49)	1.02 (-0.17 – 2.17)	<0.001
Control	312.50 ± 70.98	351.91 ± 57.47	39.41 (12.61)	0.44 (-0.83 – 1.68)	0.001
Relative peak eccentric hamstring strength (N/kg)					
	PRE	**POST**	**Mean Difference (%)**	**Hedge's *g* (95% CI)**	***p***
Nordic	4.27 ± 0.83	5.69 ± 0.79	1.42 (33.15)	1.30(0.15 – 2.41)	<0.001
Sprint	3.35 ± 0.83	4.48 ± 0.63	1.12 (33.44)	1.12 (-0.08 – 2.28)	<0.001
Control	3.14 ± 0.71	3.56 ± 0.58	0.42 (13.44)	0.47 (-0.80 – 1.72)	0.001

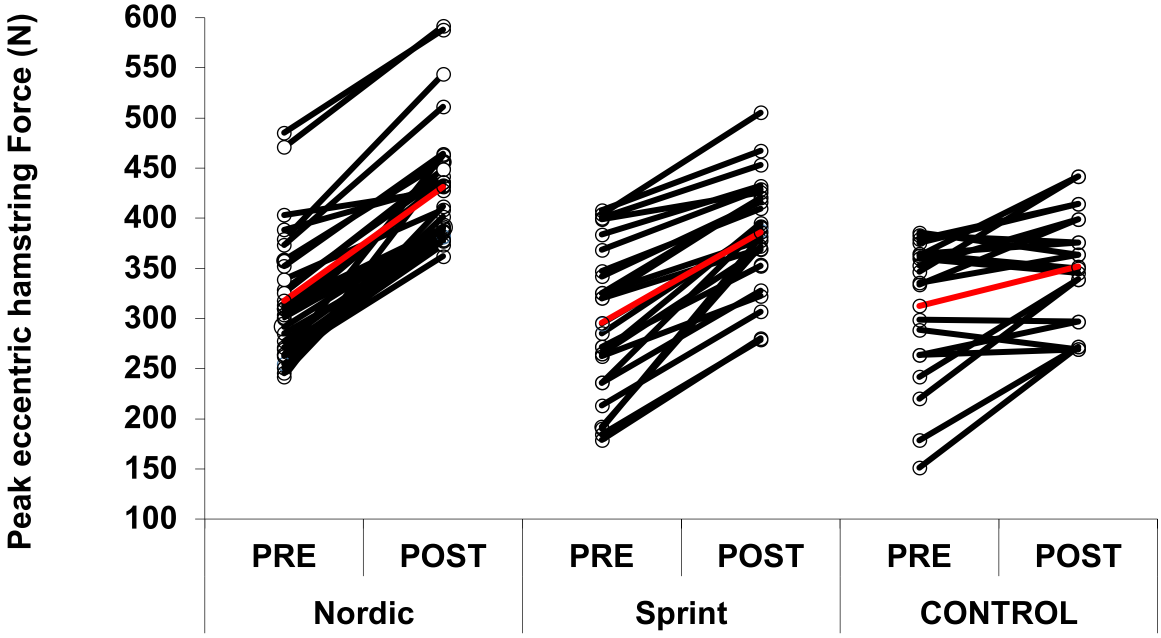


Figure 7-4 Pre- and Post-intervention individual (black) and mean (red) changes for absolute eccentric hamstring peak force.

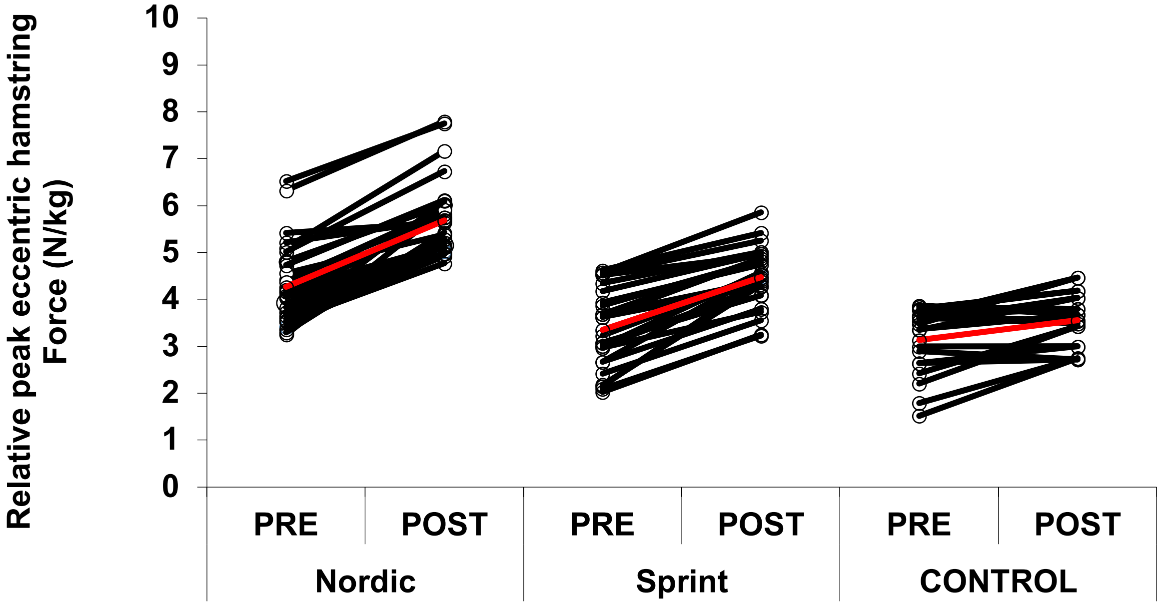


Figure 7-5 Pre- and Post-intervention individual (black) and mean (red) changes for relative eccentric hamstring peak force.

7.3.2.3 Athletic performance

For take-off velocity, non-significant time×training interaction was observed (p = 0.834). Pairwise comparisons, revealed significant and small increases were observed for both measures (Table 7-10 & Figure 7-6).

Table 7-10 Pairwise comparisons of countermovement jump measures for all training groups.

Take-off velocity ($m \cdot s^{-1}$)					
	PRE	**POST**	**Mean Difference (%)**	**Hedge's *g* (95% CI)**	***p***
Nordic	2.56 ± 0.24	2.67 ± 0.20	0.11 (4.44)	0.37 (-0.66 – 1.38)	<0.001
Sprint	2.43 ± 0.20	2.54 ± 0.16	0.11 (4.57)	0.45 (-0.66 – 1.54)	<0.001
Control	2.46 ± 0.25	2.59 ± 0.26	0.13 (5.15)	0.37 (-0.89 – 1.61)	0.001

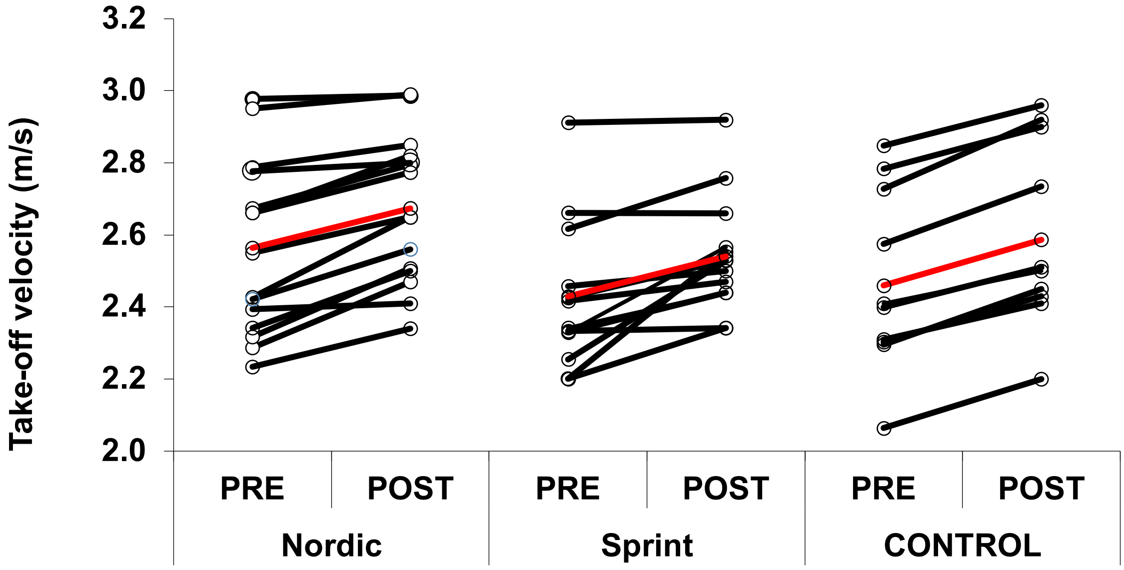


Figure 7-6 Pre- and Post-intervention individual (black) and mean (red) changes for take-off velocity.

Across the other phases of the CMJ, the countermovement phase (countermovement time, displacement, and peak braking force) showed a non-significant time×training interaction, with non-significant trivial differences from PRE-POST for all training groups. Similar findings were also reported for mean propulsion impulse with non-significant time×training interaction, with non-significant trivial differences from PRE-POST for all training groups. In contrast however, although mean propulsion force showed a non-significant time×training interaction, pair-wise comparisons revealed that the Nordic training group had a significant albeit small, increase, with non-significant and small increases for both the sprint and control training groups (Table 7-11, Figure 7-7).

Table 7-11 Pairwise comparisons of mean propulsion force for all training groups.

Mean propulsion force (N)					
	Pre	**Post**	**Mean Difference (%)**	**Hedge's *g* (95% CI)**	***p***
Nordic	1563.08 ± 288.99	1701.52 ± 279.14	132.37 (8.47)	0.37 (-0.66 – 1.38)	**0.015***
Sprint	1431.23 ± 230.81	1559.99 ± 238.00	128.75 (9.00)	0.42 (-0.69 – 1.51)	0.096
Control	1684.08 ± 286.09	1818.36 ± 263.81	134.28 (7.97)	0.36 (-0.90 – 1.60)	0.245
*** = significant increase**					

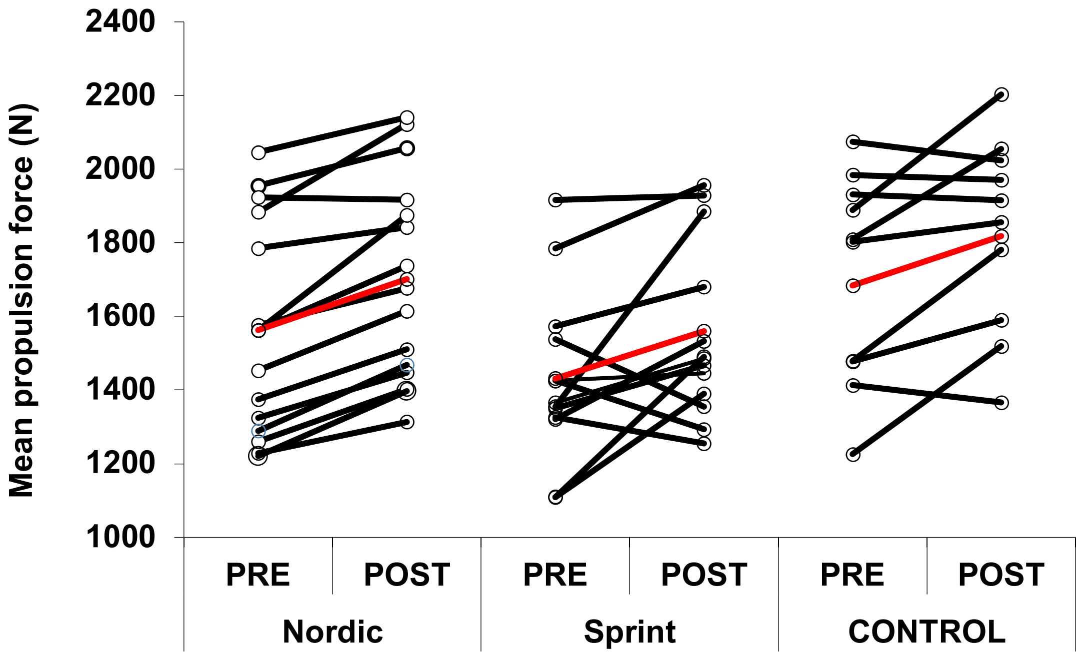


Figure 7-7 Pre- and Post-intervention individual (black) and mean (red) changes for mean propulsion force.

For peak absolute and relative net force attained from the IMTP assessment, a significant time×training interaction was observed (p = 0.013, p = 0.030) (Figure 7-8 & 7-9. Pairwise comparisons revealed that the Nordic and sprint training groups had significant and small increases in both absolute and relative peak net force (Table 7-12). The control group had a non-significant, trivial increase in absolute peak net force, although contrastingly a significant and small increase was observed for relative peak net force.

Table 7-12 Pairwise comparisons of peak net IMTP force for all training groups.

Peak absolute net force (N)					
	Pre	**Post**	**Mean Difference (%)**	**Hedge's *g* (95% CI)**	***p***
Nordic	1479.28 ± 804.67	1838.05 ± 603.80	329.64 (22.28)	0.39 (-0.64 – 1.41)	**<0.001***
Sprint	1206.78 ± 743.58	1625.74 ± 775.07	418.95 (34.71)	0.45 (-0.66 – 1.54)	**<0.001***
Control	1999.18 ± 482.45	2140.52 ± 472.64	141.35 (7.07)	0.24 (-1.01 – 1.48)	0.619
Peak relative net force (N/Kg)					
	Pre	**Post**	**Mean Difference (%)**	**Hedge's *g* (95% CI)**	***p***
Nordic	18.62 ± 9.24	23.39 ± 5.72	4.18 (22.46)	0.45 (-0.58 - 1.47)	**<0.001***
Sprint	16.72 ± 9.61	22.7 ± 9.12	5.98 (35.73)	0.48 (-0.64 - 1.57)	**<0.001***
Control	26.06 ± 4.34	28.34 ± 4.07	2.28 (8.76)	0.40 (-0.87 - 1.64)	**0.034***
*** = significant increase**					

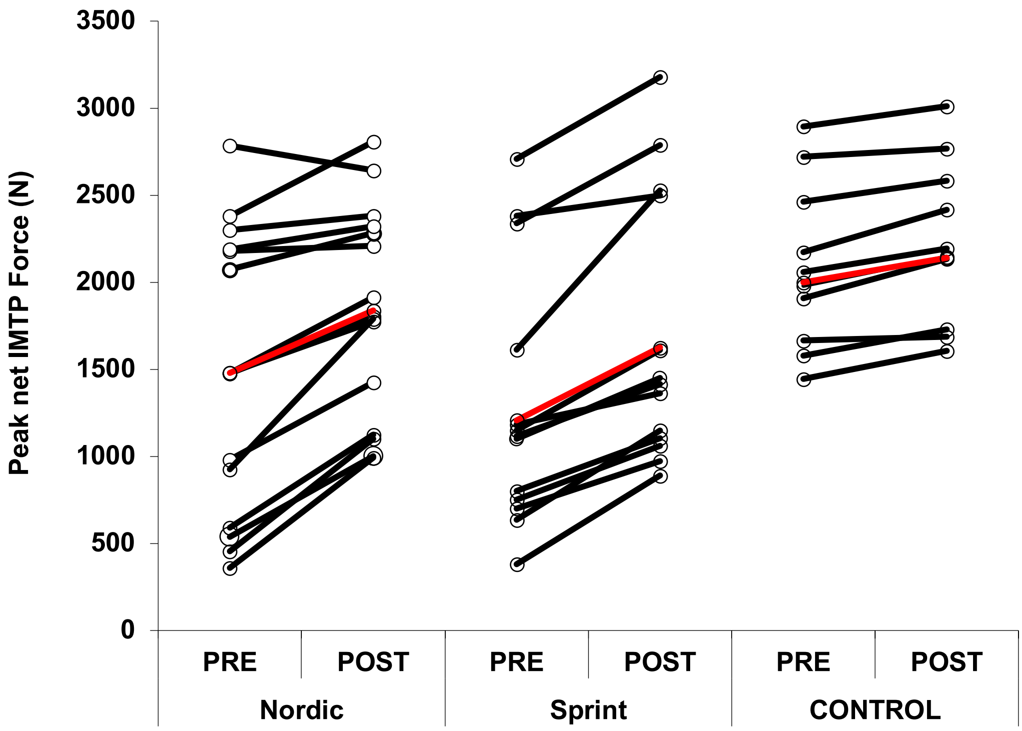


Figure 7-8 Pre- and Post-intervention individual (black) and mean (red) changes for peak net IMTP force.

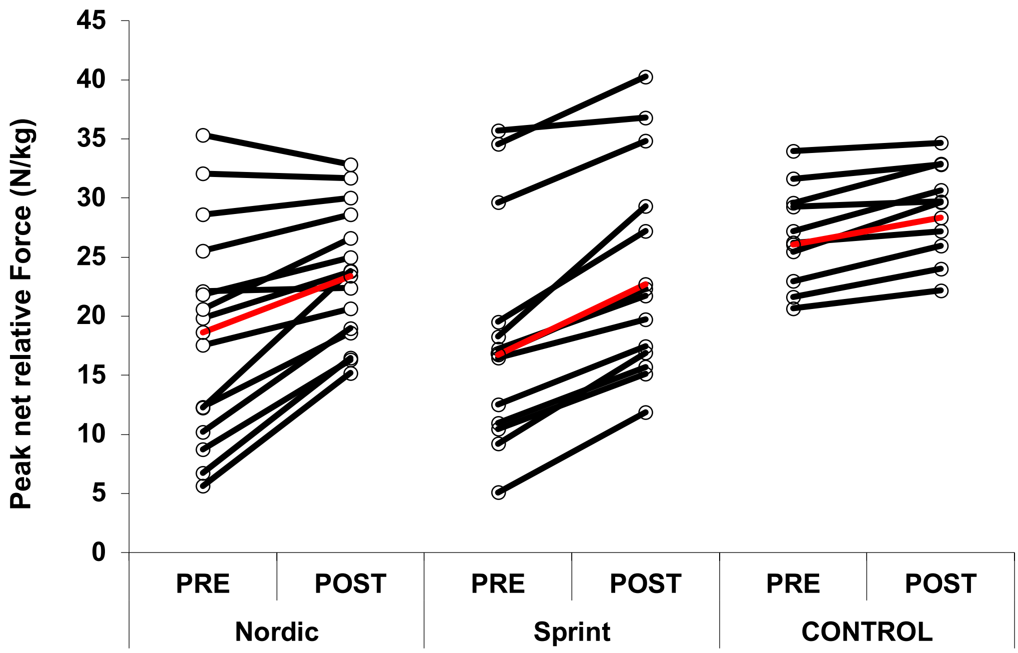


Figure 7-9 Pre- and Post-intervention individual (black) and mean (red) changes for peak net relative force.

Non-significant time×training interactions were observed for sprint and Nordic training groups for 0-10-, 0-20- and 10-20 m, *p* = 0.980, *p* = 0.699, *p* = 0.282, respectively. However, pairwise comparisons revealed significant and small decreases for both training groups for 0-10- and 0-20 m, whereas for the 10-20 m split time there were significant, small and trivial decreases for the Nordic and sprint training group, respectively (Table 7-13, Figure 7-10, 7-11 & 7-12).

Table 7-13 Pairwise comparisons of sprint measures between Nordic and Sprint training groups.

0-10m time (s)					
	Pre	**Post**	**Mean Difference (%)**	**Hedge's *g* (95% CI)**	***p***
Nordic	1.98 ± 0.13	1.90 ± 0.11	-0.08 (-4.04)	0.47 (-0.57 – 1.49)	0.001
Sprint	1.96 ± 0.11	1.88 ± 0.08	-0.08 (-4.08)	0.58 (-0.55 – 1.68)	0.002
0-20m time (s)					
Nordic	3.35 ± 0.19	3.22 ± 0.17	-0.13 (-3.88)	0.48 (-0.56 – 1.50)	<0.001
Sprint	3.34 ± 0.27	3.20 ± 0.20	-0.14 (-4.19)	0.42 (-0.69 – 1.51)	<0.001
10-20m time (s)					
Nordic	1.35 ± 0.08	1.31 ± 0.09	-0.04 (-2.96)	0.40 (-0.63 – 1.42)	0.010
Sprint	1.38 ± 0.17	1.31 ± 0.12	-0.07 (-5.07)	0.34 (-0.76 – 1.43)	<0.001

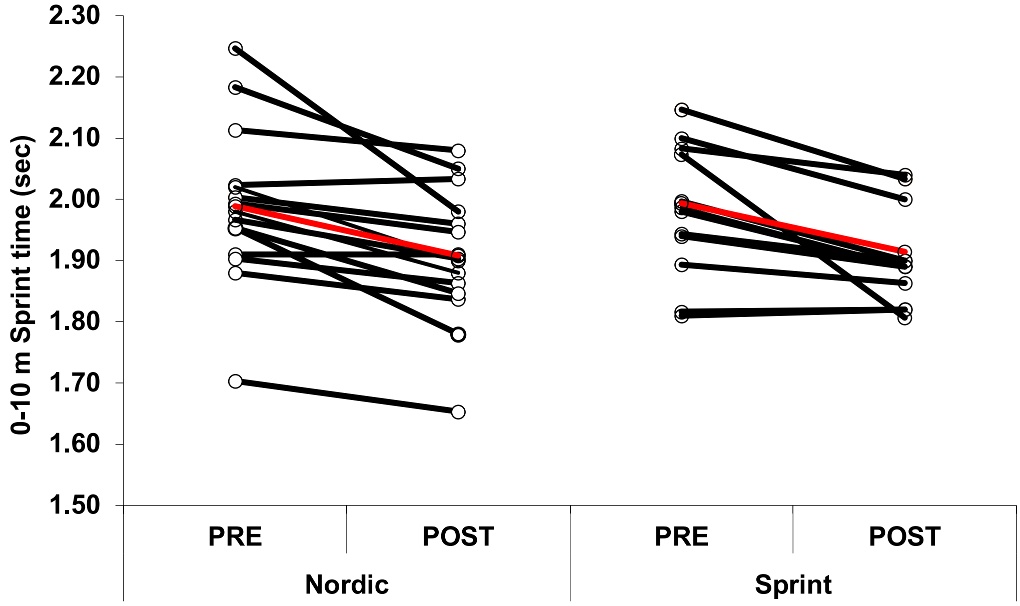


Figure 7-10 Pre- and Post-intervention individual (black) and mean (red) changes for 0-10 m sprint time.

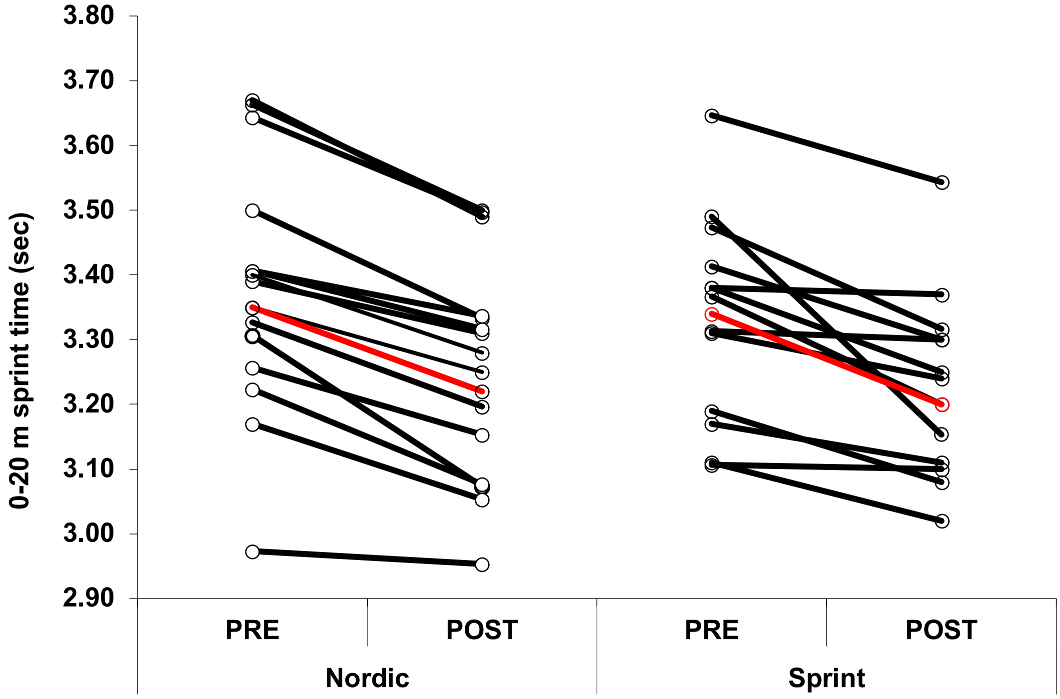


Figure 7-11 Pre- and Post-intervention individual (black) and mean (red) changes for 0-20 m sprint time.

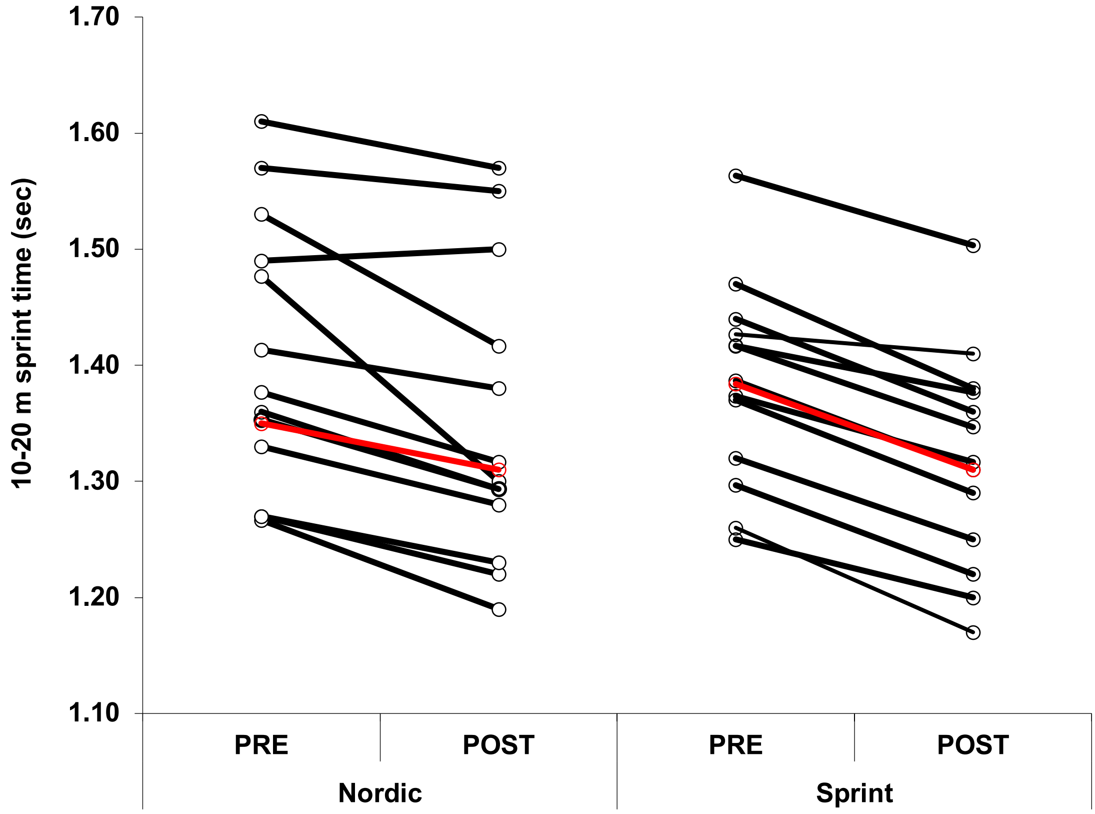


Figure 7-12 Pre- and Post-intervention individual (black) and mean (red) changes for 10-20 m sprint time.

No significant group×time interactions was observed for RPE (p = 0.964) or DOMS (p = 0.732), throughout the training intervention (Figure 7-13 & 7-14). The average RPE reported across the seven-week training period were 5.75±1.26, 5.68±0.92 and 5.68±1.37, for the NHE, sprint and control training groups, respectively.

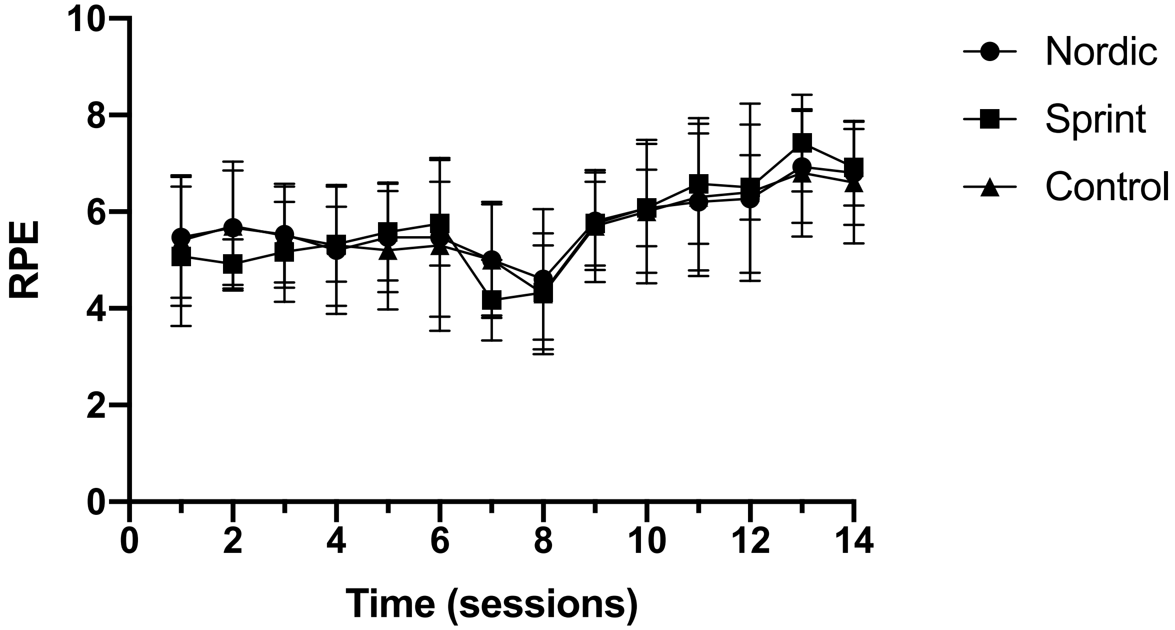


Figure 7-13 Figure 7 13 Mean (±95%CI) Rating of perceived exertion measured using a numeric scale (1–10) for the Nordic hamstring exercise, Sprint and control groups

The average DOMs reported across the seven-week training period were 3.16±1.36, 3.49±1.31 and 3.33±1.53, for the NHE, sprint and control training groups, respectively.

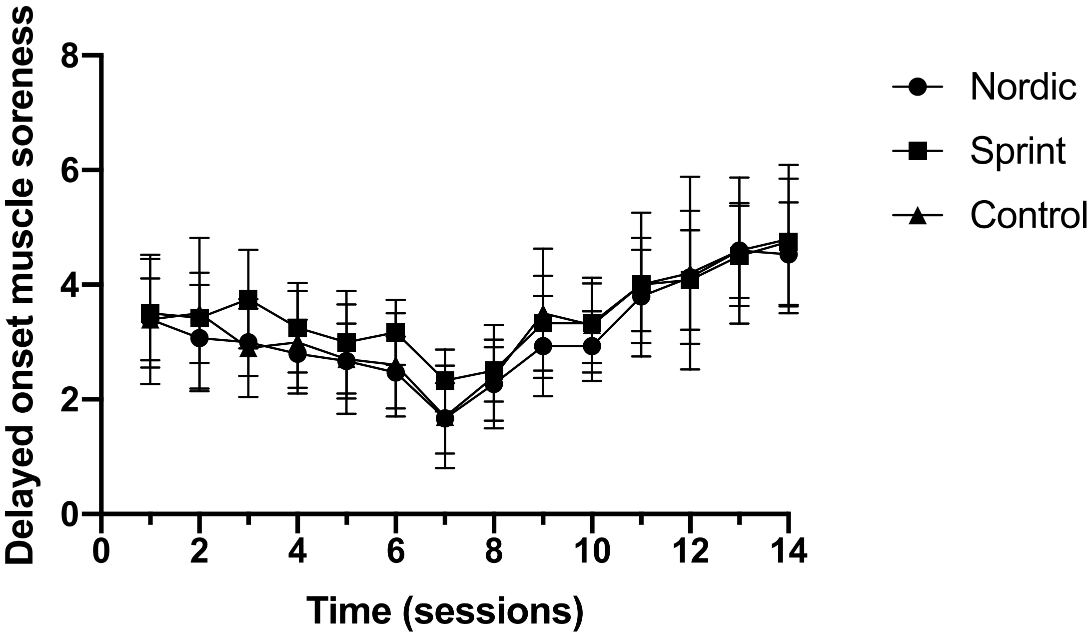


Figure 7-14 Mean (±95%CI) 24-hr post soreness measured using a numeric pain rating scale (1–10) for the Nordic hamstring exercise, Sprint and control groups

7.4 Discussion

The results of the present study demonstrate that a multi-modal approach to hamstring training is highly effective in increasing both the modifiable risk factors of HSI (eccentric hamstring strength and BF_{LH} FL), while being included within an ecologically valid training intervention that aided in increasing athletic performance. To the authors knowledge, this is the first training intervention study, that has 1) used a multi-modal approach to hamstring training (i.e., NHE + RDL, Sprint + RDL), 2) has controlled for other resistance (i.e., following a low volume, structured resistance training programme) and 3) observed multiple measure of both the modifiable risk factors of HSI and athletic performance. The results of the present study identified meaningful increases (i.e., >SDD) for the modifiable risk factors (absolute and relative BF_{LH} FL and eccentric hamstring strength) for all training groups, unsurprisingly the smallest magnitude of increase was observed within the control group – which for the present study was our single modality intervention (RDL), demonstrating the additional benefits that can be achieved with a multi-modal approach.

Similar increases in jump performance, with meaningful increases in CMJ take-off velocity observed for all training groups. The increase in take-off velocity, would also represent an increased jump height, although the smaller measurement error observed within the CMJ take-off velocity means the increases observed are less likely to be an effect of measurement error. It should be noted however, that the control group had the largest increase in CMJ take-off velocity, although the magnitude of increases was fairly similar between all groups. However, the addition of sprinting or NHE had less of an effect on jumping than the control training programme. This highlights the potential benefits to performance from hip dominant traditional exercises such as the RDL. Although in contrast, non-significant increases in mean propulsion force were observed for the sprint and control training groups it should be noted however, that all three groups had meaningful increases in mean propulsion force, to a similar magnitude, in fact the sprint training group had the greatest magnitude of increase. However, on an individual basis within the NHE training group, all bar one individual, which was within SEM, had a positive and meaningful increase within mean propulsion force. Whereas for both the sprint and control the individual response was mixed (meaningful increases and decreases in mean propulsion force). This observation for the NHE training group, indicates that using the NHE to increase eccentric hamstring strength, specifically at the knee, leads to an increased force generating capacity during hip extension (Bourne, Duhig, et al., 2017; Morin, et al., 2015). However, for all other CMJ measures assessed, there were non-significant trivial differences from PRE- to POST-intervention, this includes the countermovement phase, which was hypothesised to potentially change (increased braking force), due to the ability of the hamstrings to rapidly resist the downward motion during knee and hip flexion due to increased BF_{LH} FL (Timmins, Shield, Williams, Lorenzen, et al., 2016).

Surprisingly, the control group had non-meaningful (<SDD) increases in absolute and relative peak net force attained during the IMTP, absolute net force was trivial and non-significant increase, whereas relative net force was a small and significant increase. Both NHE and sprint intervention groups, had meaningful (>SDD), significant and small increases in both absolute and relative peak net force. The sprint training group had the largest positive increases in both absolute and relative peak net force, 34.71- and 35.73%, respectively. Followed by the NHE training group had large positive increases in both absolute and relative peak net force, 22.28- and 22.46%, respectively. The observed increases in the sprint training group could be the result of increased potential of increase motor unit activation, increase passive tension of the muscle-tendon complex and improved cross bridge mechanics (Haugen, et al., 2019). As was observed with the CMJ for NHE group, there may be an increased force generating capacity during hip extension as a result of the NHE exercise (Bourne, Duhig, et al., 2017; Morin, et al., 2015), despite the IMTP is primarily a vertical, knee extension based task – the multi-joint nature of the task with significant associations with other athletic tasks, such as sprinting, along with the ability to transfer force more effectively could explain the large positive increases for the NHE group. The non-meaningful increase within the control group was surprising as all three groups followed the same resistance training programme, therefore it was hypothesised that the same magnitude of increase would be seen for all training groups for the net force attained using the IMTP. Although, it should be noted that the control started and finished stronger than both the NHE and sprint training groups, therefore it could be expected that the magnitude of adaptations would be smaller when using any intervention for the control group (Suchomel, et al., 2016).

Unfortunately, due to track unavailability, the control group did not perform any sprint assessments. However, both the NHE and sprint training groups had meaningful and significant decreases in 0-10-, 0-20 m sprint times. Although in contrast, for the 10-20 m split time only the sprint training group reached a meaningful decrease. Across all sprint times, the sprint training group achieved the greatest decreases in comparison to the NHE training group. However, the magnitude of change did not follow this trend particularly for 0-20- and 10-20 m sprint times, potentially explained by the within-group subject variability and that the NHE were the faster group across the 0-20- and 10-20 m sprint times. Although the differences cannot be entirely attributed to the NHE or sprint training, due to the accompanying resistance training programme (Haugen, et al., 2019; Markovic, et al., 2007; Rumpf, et al., 2016), it is unsurprising that the sprint group was more effective at decreasing the longer sprint performance times such as the 0-20- and 10-20 m. However, across the shorter 0-10 m time, both groups had a near identical mean decrease (-0.08 s), the increase in sprint ability from both groups could be an effect of two different mechanisms including; greater force generating capacity during hip extension as a result of the NHE exercise (Bourne, Duhig, et al., 2017), which is specific to acceleration based tasks (Morin, et al., 2015). Along with improved structural and functioning properties of the muscle which could account for improvements in athletic performance such as; strengthened elastic properties of connective tissue, increase motor unit activation, increase passive tension of the muscle-tendon complex and improve cross bridge mechanics (Haugen, et al., 2019; Markovic, et al., 2007; Rumpf, et al., 2016).

The present study highlights novel information regarding the programming of resistance-based training for team sport athletes. Across the literature, the present study is the only study to date that has included HSI prevention, such as the NHE and sprinting, within a complete standardised training programme. The observed changes seen within the present study for BF_{LF} FL are consistent with some of previous literature (chapter 2-6 and 2-7), there are some notable differences. A lot of the decisions made with regards to the intervention study design including; training volume, progressions, duration, and exercise selection (sprint and hip dominant exercise), were made from Australian researchers (Bourne, Duhig, et al., 2017; Duhig, et al., 2019; Pollard, et al., 2019; Presland, et al., 2018), although across these studies the magnitude in changes observed were greater than those within the present study, except for body weight alone NHE prescription (Pollard, et al., 2019). However, the absolute changes in FL observed within the present study for the NHE group and those performed by the Australian research group, 1.26 cm vs 1.40-2.22 cm (Bourne, Duhig, et al., 2017; Duhig, et al., 2019; Pollard, et al., 2019; Presland, et al., 2018), could be considered similar particularly given the differences in intensity and volume, in addition to the associated error within the measurement and estimation of BF_{LH} FL that these studies employed (Franchi, Fitze, et al., 2019; Pimenta, et al., 2018). Within the present study the associated error of FL measurement and estimation was mitigated by utilising a 10- or 12-cm FOV. Contrastingly, Mendiguchia, et al. (2020) found a mean difference of 1.66 cm from a sprint training intervention, which was considerably larger than what was found within the present study for the sprint group, 0.94 cm. Despite some methodological similarities, including intervention duration and frequency of training, the exact prescription was vastly different, including both greater volumes of both sprint assistance work (i.e., resisted sprint work and plyometrics) and greater volumes of maximal effort sprints. In addition, the measurement and estimation of BF_{LF} FL was also different, which could explain the contrasting findings to the present study. Chapter 4 within the present thesis found that a 6 cm FOV, consistently overestimated BF_{LF} FL by 0.114 cm in comparison to a 10 cm FOV.

Consistent with previous training interventions (Chapter 2-7), absolute and relative eccentric hamstring strength was increased across all training groups, although with varying magnitudes. The eccentric hamstring strength changes observed for the NHE training group were larger than those highlighted within previous literature (Bourne, Duhig, et al., 2017;

Freeman, et al., 2019; Ishoi, et al., 2018; Pollard, et al., 2019; Suarez-Arrones, et al., 2019), however there are number of potential explanations as to why these studies may not have found similar changes. Firstly, Pollard, et al. (2019) and Suarez-Arrones, et al. (2019) used strong and extremely strong participants; with initial eccentric hamstring scores of 440-460 N and 570-692N for Pollard, et al. (2019) and Suarez-Arrones, et al. (2019), respectively. This indicates that the magnitude of any adaptations for the stronger athletes would be smaller across any intervention (Suchomel, et al., 2016). Across the remaining literature where the present study presented greater adaptations (Bourne, Duhig, et al., 2017; Freeman, et al., 2019; Ishoi, et al., 2018), there is the potential for methodological dissimilarities having a pronounced effect. Specifically, both Freeman, et al. (2019) and Ishoi, et al. (2018) had no progression of intensity, which is a key factor in achieving eccentric adaptation (chapter 2-7). Furthermore, despite Bourne, Duhig, et al. (2017) progressing the eccentric intensity with the addition of load – the prescription could have been excessive with high volumes. This is highlighted by Presland, et al. (2018), who used similar low session volumes that have been used within the present study, who observed large increases in eccentric hamstring strength (155 N, 2.17 *g*) (Presland, et al., 2018). However, it should be noted Presland, et al. (2018) implemented a high volume initial standardised programme, which could have resulted in the supercompensation seen in the increased eccentric hamstring strength. Applying this initial standardised programme would be impossible within practice, with the additional effect of DOMS and the potential interference with sport-based training.

With regards to other modalities used within this study (i.e. sprint and hip dominant traditional exercise), the present study found a greater change in eccentric hamstring strength than Freeman, et al. (2019). Between the present study and that of Freeman, et al. (2019) there a number of intervention design differences which could explain the greater change in the present study, firstly, the present study utilised a controlled and standardised multi-modal prescription, whereas further training was not standardised by Freeman, et al. (2019). Secondly, the short duration of the Freeman, et al. (2019) study could have influenced the POST-intervention results, as highlighted within (Chapter 2-6 and 2-7), where the longer duration interventions had the greatest positive increases. In contrast, the control group who with respect to hamstring dominant exercise only performed the RDL. There was an increase in eccentric hamstring strength, however it was only small, but a meaningful increase (39.41 N). In contrast, Bourne, Duhig, et al. (2017) using a long length traditional concentric-eccentric exercise (45° hip extension) found a large increase in eccentric hamstring strength (110.47 N). Bourne, Duhig, et al. (2017) utilised greater training volumes and intervention duration potentially explaining the difference. Furthermore, the present study capped intensity at ~75% 1RM, whereas to aid in strength development a greater relative intensity could have been prescribed, more in line with strength training recommendations (Sheppard and Triplett, 2016).

The present study was highly effective at increasing both modifiable risk factors of HSI (eccentric hamstring strength of BF_{LH} FL), as well as increasing athletic performance. One potential explanation as to why this study had such positive effect was that it achieved 100% compliance, potentially due to the low volume approach adopted as part of the intervention, similar to what would be performed in-season within sport. The low volume approach utilised within the present study also limited the effect of DOMs with only moderate DOMS and RPE reported (Figure 7-13 and 7-14), even as participants were progressed up to higher eccentric intensities. Notably, the individual DOMs ratings did not decrease during the intervention period, contrasting much of the literature regarding repeated bout effect and eccentric training (Howatson and van Someren, 2007; Mchugh, 2014). This observation could be due to the DOMs rating being of a total body soreness rather than specifically to the hamstrings, which was thought to be more relevant to practice and sport. Although it should also be mentioned that the NSCA foundation grant will also have had a positive effect on compliance, as the subjects were paid for their participation to achieve 100% compliance. A minimum of 75% compliance was demonstrated to have most positive beneficial effect of HSI incidence, and it would be suspected that a similar finding would be observed for the modifiable risk factors of HSI (Bourne, et al., 2018). With regards to application, a low volume approach to the NHE and sprinting used within practice could achieve greater volumes of compliance, specifically as a low volume NHE appears to have minimal influence DOMs. Additionally, sprinting can be made competitive, with immediate feedback further enhancing the positive experience that athletes can have when performing sprint training, increasing athlete compliance, with high levels of compliance (>80%) observed previously for sprint training (Mendiguchia, et al., 2020). However, a similar intervention using bounding could not achieve high levels of compliance, failing to reach the 75% identified, achieving a moderate level of compliance of 71% where a bounding exercise programme did not prevent HSI incidence (Van de Hoef et al., 2017).

The positive findings with regards to the modifiable risk factors of HSI within the present study, highlight both NHE, sprint training as well as the RDL, could have positive effects upon HSI incidence. Moreover, the combination of methods was more effective than the control group (long length traditional concentric eccentric exercise only) as well as more effective than some of the previous literature that has only utilised a single modality, indicating that a multi-modal approach may be more effective at decreasing rate of HSI within sport. To date the NHE has been shown to be the most effective exercise that we know of that decreases HSI incidence (Chapter 2-5), although compliance is key influencer on any modality's effectiveness. Whereas similar observations have currently not been conducted with sprint-based or long length traditional concentric eccentric exercise interventions. To date, the closest investigation to sprinting was performed by a Danish research group (Van de Hoef, et al., 2017), where the effect of a bounding exercise programme on HSI occurrence within soccer players was observed within a large, randomised control trial. They found no evidence that a bounding exercise programme prevented HSI occurrence (Van de Hoef, et al., 2017), with large spikes in the total work performed possibly explaining why the bounding plyometric exercise programme did not reduce HSI occurrence; and may have indirectly increased the risk of future HSI incidence. Although no measurements of hamstring strength or muscular architecture were made, it could also be presumed that in contrast to maximal sprint running, bounding exercises did not induce similar magnitudes of hamstring loading to positively increase eccentric hamstring strength of BF_{LH} FL, due to a reduced movement velocity, with a focus on joint stiffness. However, as suggested a single modality would be very uncommon within practice, even the present study is the first to observe the effect of a controlled multi-modal intervention, therefore, it would be more relevant to practice for future research to observe the effect of multi-modal interventions upon the modifiable risk factors of HSI and HSI incidence, although the latter would come with extreme difficulty.

The present study is not without its limitations; firstly, although all participants reported participation in regular sport (predominantly team sport); competitive level, season, positional demands were not collated, therefore the influence of the sport demands could not be classified between any groups. This meant that individuals would have been exposed to a variety of external running loads and training loads, which could have all influenced the individual responses observed during the intervention (Freeman, et al., 2019; Timmins, et al., 2017). Despite the non-standardised nature of external training, both eccentric hamstring strength and BF_{LH} FL saw increases across the sample, with individual for every participant. Although, this also highlights a strength of the present study as it ecologically valid, as it based within a complete resistance training programme, where individuals were still participating within sport. A further limitation of the present study was that the assessment of eccentric hamstring strength was made using the NHE, i.e., training for the test rather than the potential adaptation. Especially as there is limited agreement between the Nordbord and isokinetic methods of hamstring assessment (Wiesinger, et al., 2019). It would have been prudent to assess eccentric hamstring strength using an isokinetic eccentric assessment, to have a more comprehensive understanding of the eccentric adaptations to the training program. This type of method was not employed due time and availability of equipment, specifically the time to assess individual participants using isokinetic assessments.

The application of the training intervention could have been improved with appropriate feedback or technical modification. Real time visual feedback has been previously shown to increase peak mean eccentric peak force in the NHE within athletes (Chalker, et al., 2018), with suggestions that this could improve the adaptive response. Therefore, over the extended period used within the present study, the use of augmented real time feedback could have resulted in even greater adaptations the NHE training group, than those presently observed. Additionally, the sprint training groups' application could have been improved by the utilisation of various drills and video feedback (Figure 8-2) which could aid in technical modification. Although some of this may have added to overall training volume (i.e., distance), it could enhance the technical proficiency of participants potentially having a greater positive effect upon observed adaptations.

Finally, seven of the participants were not able to attend testing at the University of Salford and therefore testing took place off-site, while the majority of the methodologies were able to remain consistent using the Nordbord device and force platforms for tests of eccentric hamstring strength and athletic performance. For the assessment of BF_{LH} FL, as the large single FOV US device was not transportable, therefore, a portable device with a shorter probe was utilised. The method chosen to assess with the shorter FOV maximised the analysis by using an extended FOV method, doubling the image along the line of the muscle (6-cm + 6-cm) to achieve an overall FOV of 12-cm. Sprint assessments and sprint training also took place on a 3G AstroTurf, despite the participants who were tested off site being prepared and accustomed to performing sprint running on an AstroTurf as part of their regular training and wearing appropriate footwear. The difference in surface could influence the results observed within both testing results and training adaptations, therefore this is another noteworthy limitation. Furthermore, due to track unavailability at the University of Salford, the control group was not able to perform any sprint assessments, this means that the conclusions made about the effect of sprint and NHE training upon improvement in sprint ability should be taken with caution. As the effect of the standardised training programme were not identified, as it would be expected increases in strength (i.e., IMTP peak net force), through the periodized resistance training programme would also transfer to sprint performance.

7.5 Conclusion

The present chapter set out to determine the effect of a short-term training intervention with supplemental sprint or NHE, imbedded within an ecologically valid training programme, on the magnitude of adaptations to the modifiable risk factors of HSI, i.e., BF_{LH} muscle architecture and eccentric hamstring strength, and athletic performance. The findings suggest in general that utilising the NHE within an ecologically valid training programme results in meaningful increases in BF_{LH} FL and eccentric hamstring strength, to a greater magnitude than sprinting and the resistance training programme containing the RDL alone. Further inspection demonstrated that on an individual level all participants from each group increased BF_{LH} FL and eccentric hamstring strength, this evidence indicates that all training methods used within the present study increases in BF_{LH} FL and eccentric hamstring strength, supporting the findings of Chapters 2-6 and 2-7 that identified that across all training methods utilised within the literature, there is a resultant increase in eccentric hamstring strength. However, a multi-modal approach to training does have the greatest positive effect upon modifiable risk factors of HSI. This is an important practical application for strength and conditioning coaches, sports rehabilitators and sport scientists, in that HSI prevention should not come in a single form – it should form part a multimodal prescription containing multiple elements (e.g., NHE, sprinting and hip dominant concentric-eccentric). A natural progression of this work is to observe the effect of these types of interventions on HSI incidence with sport. Although it the author suggests that before performing interventions to observe the effect on HSI incidence within sport, a study similar to the present one should be carried out in an attempt to understand how to optimise sprint-based training, specifically optimal volumes, frequencies and modalities which all have a positive effect on increasing BF_{LH} FL and eccentric hamstring strength.

8 Thesis Summary and Recommendations for Future Research

8.1 Summary and Conclusions

The overarching aim of the thesis was to be able to inform the utilisation of a cyclical-practice format (assessment, performance and training) (Figure 8-1), beginning with the identification of appropriate methods to assess two of the primary modifiable risk factors to HSI (eccentric strength and fascicle length), followed by identifying how eccentric strength and fascicle length influence running characteristics and subsequently how specific training may mitigate the risk of HSI via the adaptations to eccentric strength and fascicle length. The overarching aim and the underlying objectives of the present thesis addresses what occurs within informed practice, where there is an initial assessment, with subsequent follow ups, including observing any effect on performance and adaptations through training.

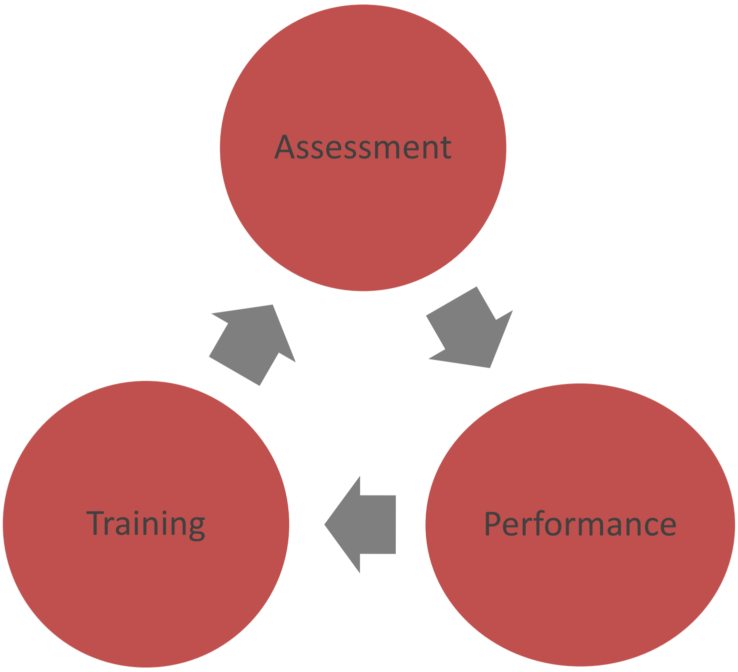


Figure 8-1 Informed cyclical-practice format to optimise athletic performance and development of a robust approach to HSI risk reduction.

To date, the assessment of eccentric hamstring strength has been fairly well standardised, with both a gold standard, lab-based assessment and field-based measures accepted. However, there remained a number of unanswered questions regarding the assessment of BF_{LH} FL. The author of the present thesis was able to conclude that using a 10-cm FOV to assess BF_{LH} muscle architecture via ultrasonography is highly reliable, regardless of the estimation equation utilised. Furthermore, both the estimation equation and FOV utilised are highly influential factors; this is a key finding as research and practice commonly utilises a short FOV (<6cm), with previous literature comparing to more time consuming, less practice friendly techniques (Franchi, Fitze, et al., 2019). Equation 3-3 was found to be the most accurate as it removes a large degree of the required estimation, therefore, it was utilised in the remainder of the thesis.

Following on from the assessment phase, the influence of eccentric hamstring strength and BF_{LH} FL on running performance was identified. Upon allocating individuals to either a low or high-risk group, by using the nearly perfect relationship between relative measures of eccentric hamstring strength and BF_{LH} FL, there were meaningful differences on both peak and waveform lower-limb kinematics and neuromuscular contributions of the hamstring muscles during treadmill running. The kinematic and neuromuscular characteristics observed within individuals with a shorter relative BF_{LH} FL and lower eccentric hamstring strength, could heighten the risk of HSI occurrence. This is due to higher risk group demonstrating a greater magnitude and rate of muscle length change, a greater change in the pelvis co-ordination (pelvic tilt) and a greater degree of movement variability, in addition to a greater neuromuscular contribution of the BF_{LH}. All the highlighted differences may be contributing factors to HSI incidence during running (Chumanov, et al., 2011; Heiderscheit, et al., 2005; Opar, et al., 2012; Schache, et al., 2012; Shield and Bourne, 2018; B. Yu et al., 2008).

With respect to training, altering the performance angle of the NHE had significant and meaningful effect. Specifically, when the NHE was performed at an increased angle there was a greater knee angle and MTU length at break point, in addition to a shift from a medial to lateral neuromuscular contribution of the hamstrings. However, there were non-significant and trivial differences in knee angle at break point relative to the horizontal, which was used as a simple measure for the moment occurring at the knee, as well as the neuromuscular measures of both the lateral and medial components. Furthermore, the observations of BF_{LH} *in vivo* muscle mechanics, found meaningful differences within the early-mid range of movement (0-40% time), where greater fascicle shortening was observed within the decline and flat NHE variations. However, there was no likely or meaningful differences identified between variations at the mid-end range (40-100% time). By utilising the incline NHE variation within a training programme, especially within the early phases of training, it could potentially provide a stronger positive adaptation to the BF_{LH} architecture and eccentric hamstring strength as at the instance of eccentric overload i.e., breakpoint, the hamstring complex would be working at a greater muscle length with a greater neuromuscular contribution of the lateral component. Despite not being a greater magnitude of expected force, the fact the muscles would be working at a greater length is highly influential in the adaptive response especially in eccentric training (Guex, Degache, et al., 2016; Sarabon, et al., 2019). However, the *in vivo* muscle mechanics and knee at break point to the horizontal measurements would suggest that there may be minimal difference between the variations, with a similar magnitude of fascicle lengthening occurring at the mid- to end-range of movement and the moment achieved across variations.

A holistic approach to the training intervention was performed to support its ecological validity, including using two multi-modal groups (NHE + RDL and sprint + RDL), in addition to a unimodal group used as a control (RDL only). Across all training groups, increases in BF_{LH} FL and eccentric hamstring strength were observed, albeit to varying magnitudes. The NHE + RDL training group demonstrated the greatest magnitude of adaptation in BF_{LH} FL and eccentric hamstring strength, which was subsequently followed by the sprinting group, and then the control group. To no surprise, this highlights that multi-modal practice, or a holistic approach is superiorly effective than unimodal, and therefore hamstring exercises should not be used in isolation. Moreover, this was performed as part of a complete, albeit simple and low volume, lower body training programme that was able to elicit increases in athletic performance. Although the increases in athletic performance, cannot be directly attributed to the increases in eccentric hamstring strength or BF_{LH} FL, via the application of the NHE, sprinting or RDL, as per previous literature (Askling, et al., 2003a; Clark, et al., 2005; Ishoi, et al., 2018; Krommes, et al., 2017), it does highlight that a complete training programme incorporating these exercises can have a positive and meaningful influence on athletic performance and primary modifiable risk factors of HSI.

8.2 Limitations and recommendations

The present thesis is not without its limitations. Firstly, it could be suggested that the link between the individual studies is somewhat tenuous and potentially did not go as far, leaving further areas that could be addressed or studied. For instance, chapter 7 did not include the identification of any changes in the lower limb kinematics during running from pre- to post-training, due logistical challenges. This is despite chapter 5 highlighting that individuals with low eccentric hamstring strength and BF_{LH} FL displayed altered running kinematics and recent research identifying that using the NHE has a positive effect on swing phase mechanics (Alt, et al., 2021). Within chapter 7 there was a large quantity of data that was planned to be collected, this included various measures of athletic performance, which was required as if a training intervention is detrimental to athletic performance, it cannot be deemed an ecological success. Due to the large quantity of data being collected, in addition to the restricted time to collect data within participants (due to scheduling conflicts at the University and the team sport environment) a decision was made to not perform any kinematic analysis of sprint running and change the measure of strength from the gold standard lab based isokinetic dynamometry to the field based Nordbord assessment, which is an appropriate and effective more time efficient measure of eccentric hamstring strength, despite differences between the two methods (Wiesinger, et al., 2019). This could also explain why the NHE group had the greatest increase in eccentric hamstring strength, as they were regularly performing the NHE. Although measures such as time to peak force and active impulse metrics, which could differentiate between NHE strategies, were observed, the changes were non-meaningful different between training groups (Appendix Four). Therefore, despite the studies perhaps not connecting as well as they could have done, practical applications and recommendations for future are provided in each study to inform how these gaps may be addressed in future studies.

Following on from the above, determining changes in lower limb kinematics during running, would have been a beneficial addition to determining the effectiveness of training intervention (chapter 7), especially with respect to the observations made in chapter 5 and the differences between high- and low-risk groups. This would have provided some level validity to the observations of Chapter 5, especially if positive meaningful changes were observed in eccentric hamstring strength and BF_{LH} FL in conjunction to positive changes to running kinematics. However, to complete this process would have required either 3D motion capture, which again would have been a time-consuming process, or 2D motion capture but this would first require validation and comparison to the differences observed using 3D motion capture. Therefore, this does leave an opportunity for a future progression of the research, not only in terms of validation between 2D and 3D motion capture to observe differences in running kinematics (i.e., greater knee extension and extension velocity during the terminal swing phase, greater “backside” running mechanics), but also regarding whether changes in eccentric hamstring strength and BF_{LH} FL (as a result of resistance or sprint based training) can provide positive changes in running kinematics identifiable during 2D motion analysis. Although this is not novel or ground-breaking practice, validating it as a tool that could indicate both limitations in technique and eccentric hamstring strength and/or BF_{LH} FL would be highly beneficial (Josse, 2020; McMillan and Pfaff, 2018). Recently, this type of analysis has been promoted by sprint coaches in the provision of technique analysis and feedback, specifically the Altis Kinogram (Figure 8-1) (Josse, 2020; McMillan and Pfaff, 2018) or other alternative 2D methods including the 'kick-back' mechanism (Figure 8-1) (Lahti, et al., 2020).

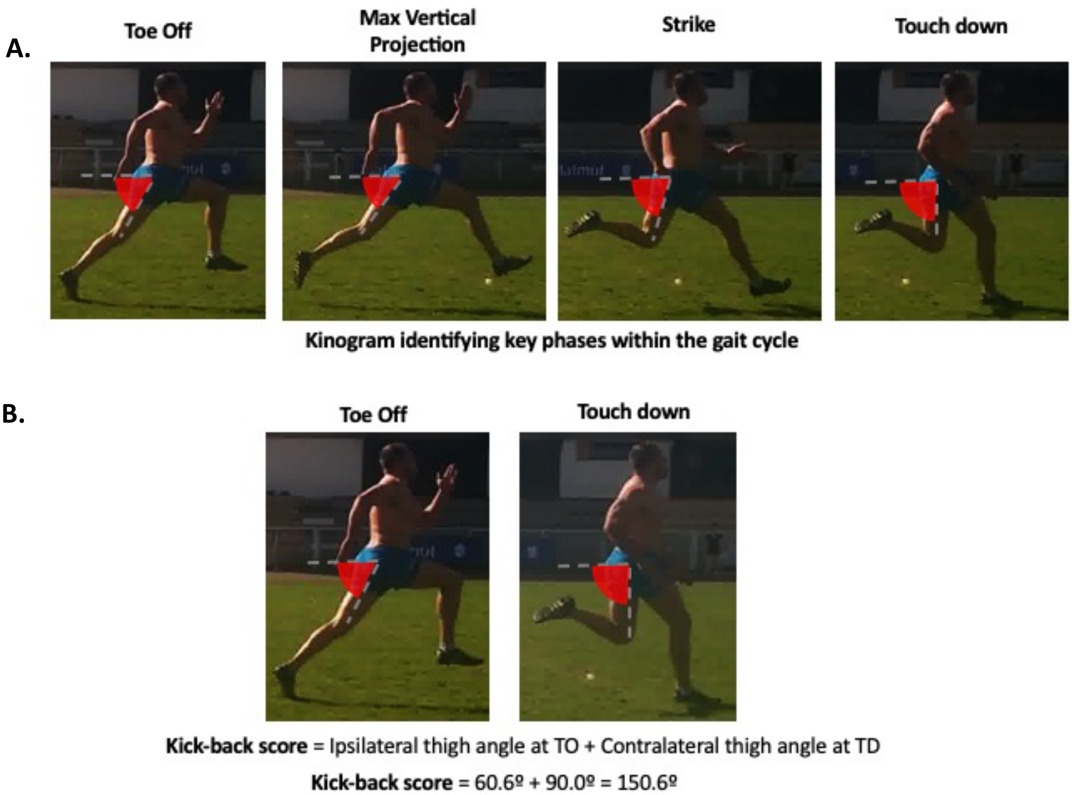


Figure 8-2 Two methods of 2D analysis of lower limb kinematics during sprint A. Kinogram, identifying key phases within the gait cycle potentially used for qualitative or quantitative analysis (Josse, 2020; McMillan and Pfaff, 2018). B. Kick-back mechanism identifying optimal or sub-optimal kick-back (Lahti, et al., 2020).

The findings of Chapter 6 highlighted that NHE variations could be used a potential natural progression within training, as the observations in kinematics, EMG and dynamic US indicated that the adaptations from an incline NHE maybe superiorly effective, especially for relatively novice athletes who would then be able to progress more appropriately to flat and decline variations. However, the consistency in recommendations (volume, intensity, and progression) within the current literature around training prescription to increase BF_{LH} FL and eccentric hamstring strength is limited, especially with regards to elite practise. Therefore, to complete the research aim and underlying objective of a more ecologically valid approach, it was decided that a known NHE prescription would be utilised for training phase of this thesis. This included performing the NHE across a standardised performance angle (flat), with a low volume, progressive intensity, in line with previous recommendations (Bourne, Duhig, et al., 2017; Pollard, et al., 2019; Presland, et al., 2018; van Dyk, et al., 2019). Future research could therefore look to establish what the effect of training across different NHE performance angles is on BF_{LH} FL and eccentric hamstring strength, thus determining if it is a logical progression within practice as suggested.

Across the thesis there was a change in sample population, within chapters 3-5 the sample was only male, this was through no criteria but just a consequence of convenience sampling. However, for the final two chapters a mixed cohort of male and female participants were included and as the menstrual cycle was not accounted for within either chapters for the female participants, this may have influenced the outcome measures, as the menstrual cycle has been reported to have negative effects upon both strength and anaerobic performance (Carmichael, Thomson, Moran, & Wycherley, 2021). However, this does not detract from the findings with regards to training for positive adaptations in the modifiable risk factors, as female athletes still sustain HSIs, although not as frequently as male counterparts. Additionally, the adaptations to the modifiable risk factors could also benefit female athletes by reducing the risk of sustaining and ACL injury, which occurs more frequently in females.

8.3 Practical applications

Assuming that athletes are injury free, with no recent history of HSI; US assessment of BF_{LH} architecture can be performed reliably using a 10cm FOV and any of the estimation equations identified. Within the present thesis, SEM and SDD values were also established, as this is the first instance of using a 10cm FOV to assess BF_{LH} architecture. The measurements of the BF_{LH} architecture should not be used interchangeably between different estimation equations and FOV, it is suggested that practitioners should choose a single method and use it consistently. It is recommended that practitioners utilise the partial measure equation (equation 3-3), as this method reduces the degree of estimation required in comparison to the alternative equations.

The implementation of the NHE has typically involved performance in a horizontal plane, however, the performance angle has a meaningful impact upon kinematics around the break point of the NHE, suggesting altered torque-angle curves, changes to the BF_{LH} *in-vivo* muscle mechanics and a lateral shift in the neuromuscular response of the hamstrings. Therefore, altering the NHE performance angle could be an appropriate progression or regression for athletes, specifically those with limited equipment, potentially leading to greater meaningful increases in the HSI modifiable risk factors than conventional methods, due to the differences in kinematics, neuromuscular response, and *in-vivo* muscle mechanics. Additionally, the Hamstring Solo (ND Sports performance, Thomastown, Ireland), a novel testing and training device similar to the Nordbord, is angled at a decline. Not only will this influence performance differences between the Nordbord and the Hamstring Solo, but it could also negatively affect the NHE performance, in comparison to neutral angle. This is crucial information as if athletes are only performing the NHE on the Hamstring Solo, then the adaptive response could be blunted or delayed.

Practitioners can use the information within the present thesis to make informed decisions with regards to injury prevention practices. A variety of exercises were utilised, with the NHE following low volume, progressive intensity prescription, a low but progressive volume of sprint running intervention and moderate volume and intensity RDL, providing a sufficient training stimulus to have a positive and meaningful effect upon the modifiable risk factors of HSI. Some of these exercises and exercise prescriptions are novel within the literature, with a focus on being practice friendly where athletes were still required to regularly perform their individual sport along with an entire lower body resistance training session. Elite team sport practice is typically not focused on a single modality or single element, with a multitude of physical elements being brought together to optimise performance and minimise the risk of injury. In support of this statement, the present thesis highlights that a multi-modal intervention is the most effective in increasing athletic performance and reducing the potential risk of HSI by increases in eccentric hamstring strength and BF_{LH} FL. It is hoped this information can aid practitioners in forming decisions upon a holistic, prophylactic training programme within sport.

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Appendices

Appendices One – Hamstring strain injury occurrence observations

Table Appendix One - Hamstring strain injury occurrence across sports.

Study or report	Sport	Gender	Observation period	Age	n	Reported rate calculation				
						/1000 player hours	/season	/total exposures	/1000 athletes	/10,000 exposures
AFL Injury survey (2017)	AFL	Male	1997 - 2017	18+	N/A		6.00			
Verrall, Slavotinek and Barnes (2005)		Male	Pre-2005	18+	169	2.13				
Edouard, Branco and Alonso (2016)	Athletics	Mix	2007 - 2015	18+	N/A				22.40	
Camp et al, (2018)	Baseball	Male	2011-2016	18+	N/A		556.10			
Longo et al, (2012)	Basketball	Male	2009 - 2010	15.2	41	0.00				
Orchard et al., (2010)	Cricket	Male	1998 - 2009	26.1	33		3.90			
Roe et al, (2018)	Gaelic football	Male	2008 - 2015	18+	307	2.20				
Dalton, Kerr and Dompier (2015)	NCAA athletes	Mix	2009 - 2014	18-25	N/A					3.05
England professional Rugby injury surveillance project (2016)	Rugby	Male	2015 - 2016	18+	N/A		4.01			
Tee, Till and Jones (2018)		Male	2017	17.8	81	2.00				
Beaudouin et al, (2018)	Soccer	Male	2014 - 2015	11.3	1829	0.01				
Cross et al, (2013)		Mix	2004 - 2009	18-23	N/A	0.59				
Engebretsen et al, (2008)		Male	2004	17 - 35	195			0.90		
Ekstrand, Hagglund and Walden (2011)		Male	2001 - 2009	25.3	4658	2.48				
grooms et al, (2013)		Male	2010	18-25	30	3.12				
Junge et al, (2004)		Male	2001	16.4	145	4.70				
Petersen et al, (2011)		Male	2008 - 2009	23.5	481	13.10				
Silvers-granelli et al, (2015)		Male	2012	18-25	850			1.24		
Soligard et al, (2009)		Female	2007	13-17	837	0.20				
van der Horst et al, (2015)		Male	2013	18-40	287			0.90		
Whalan et al, (2018)		Male	2016	24.3	1049	3.00				
Valle et al, (2018)	Youth team sports	Male	2007 - 2010	13.96	1157	0.15				

Appendices Two – Between group stride variability across kinematic measures and running velocities

Table Appendix Two Between group stride variability for kinematic measures of the left limb.

		8 km·hr^{-1}		10 km·hr^{-1}		12 km·hr^{-1}		14 km·hr^{-1}		16 km·hr^{-1}	
		HIGH	**LOW**	**HIGH**	**LOW**	**HIGH**	**LOW**	**HIGH**	**LOW**	**HIGH**	**LOW**
LEFT	**Peak Hip Extension**	0.788 (0.667 - 0.895)	0.991 (0.975 - 0.997)	0.798 (0.694 - 0.899)	0.992 (0.979 - 0.997)	0.798 (0.695 - 0.899)	0.998 (0.993 - 0.999)	0.798 (0.694 - 0.899)	0.994 (0.985 - 0.998)	0.792 (0.678 - 0.897)	0.996 (0.988 - 0.998)
	Peak Hip Flexion	0.655 (0.273 - 0.856	0.699 (0.347 - 0.877)	0.370 (0.291 - 0.662)	0.587 (0.168 - 0.824)	0.420 (0.150 - 0.570)	0.778 (0.672 - 0.950)	0.285 (0.108 - 0.450)	0.414 (0.263 - 0.671)	0.384 (0.098 - 0.619)	0.528 (0.084 - 0.795)
	Peak Knee Extension	0.791 (0.675 - 0.897)	0.990 (0.972 - 0.996)	0.795 (0.687 - 0.898)	0.993 (0.981 - 0.997)	0.795 (0.685 - 0.898)	0.991 (0.975 - 0.997)	0.797 (0.693 - 0.899)	0.997 (0.992 - 0.999)	0.796 (0.688 - 0.898)	0.995 (0.986 - 0.998)
	Peak Knee Flexion	0.724 (0.390 - 0.888)	0.768 (0.471 - 0.907)	0.544 (0.114 - 0.805)	0.706 (0.556 - 861)	0.751 (0.504 - 0.955)	0.917 (0.790 - 0.969)	0.847 (0.630 - 0.940)	0.886 (0.716 - 0.956)	0.249 (0.016 - 0.412)	0.498 (0.043 - 0.779)
	Peak Change in Knee Angular Velocity	0.352 (0.128 - 0.574)	0.601 (0.196 - 0.833)	0.402 (0.205 - 0.607)	0.641 (0.251 - 0.850)	0.301 (0.199 - 0.404)	0.613 (0.208 - 0.845)	0.341 (0.147 - 0.535)	0.555 (0.121 - 0.808)	0.759 (0.454 - 0.903)	0.780 (0.327 - 0.973)
	Peak Bicep femoris muscle tendon unit length	0.746 (0.431 - 0.898)	0.747 (0.515 - 0.962)	0.742 (0.422 - 0.896)	0.732 (0.553 - 0.912)	0.459 (0.227 - 0.674)	0.706 (0.391 - 0.858)	0.745 (0.428 - 0.897)	0.887 (0.717 - 0.956)	0.746 (0.430 - 0.898)	0.746 (0.430 - 0.898)
	Take off (% gait)	0.988 (0.968 - 0.996)	0.988 (0.967 - 0.995)	0.972 (0.925 - 0.990)	0.923 (0.803 - 0.971)	0.978 (0.942 - 0.992)	0.960 (0.893 - 0.985)	0.982 (0.951 - 0.993)	0.958 (0.888 - 0.984)	0.943 (0.851 - 0.978)	0.974 (0.930 - 0.990)

Table Appendix Two Between group stride variability for kinematic measures of the Right limb.

		8 km·hr^{-1}		10 km·hr^{-1}		12 km·hr^{-1}		14 km·hr^{-1}		16 km·hr^{-1}	
		HIGH	**LOW**	**HIGH**	**LOW**	**HIGH**	**LOW**	**HIGH**	**LOW**	**HIGH**	**LOW**
RIGHT	**Peak Hip Extension**	0.786 (0.662 - 0.895)	0.993 (0.981 - 0.997)	0.798 (0.695 - 0.899)	0.995 (0.987 - 0.998)	0.797 (0.692 - 0.899)	0.997 (0.992 - 0.999)	0.797 (0.692 - 0.899)	0.996 (0.988 - 0.998)	0.795 (0.687 - 0.898)	0.998 (0.005 - 0.999)
	Peak Hip Flexion	0.631 (0.235 - 0.845)	0.779 (0.492 - 0.912)	0.547 (0.109 - 0.804)	0.798 (0.529 - 0.920)	0.664 (0.288 - 0.861)	0.671 (0.299 - 0.864)	0.401 (0.246 - 0.647)	0.536 (0.451 - 0.960)	0.544 (0.106 - 0.803)	0.789 (0.512 - 0.916)
	Peak Knee Extension	0.789 (0.671 - 0.896)	0.986 (0.961 - 0.995)	0.784 (0.658 - 0.894)	0.989 (0.971 - 0.996)	0.982 (0.653 - 0.893)	0.983 (0.953 - 0.994)	0.794 (0.683 - 0.898)	0.992 (0.977 - 0.997)	0.788 (0.667 - 0.896)	0.992 (0.977 - 0.997)
	Peak Knee Flexion	0.380 (0.103 - 0.612)	0.745 (0.429 - 0.897)	0.407 (0.189 - 0.596)	0.417 (0.158 - 0.575)	0.483 (0.187 - 0.670)	0.614 (0.470 - 0.758)	0.180 (0.123 - 0.237)	0.252 (0.014 - 0.442)	0.362 (0.580 - 0.704)	0.421 (0.054 - 0.684)
	Peak Change in Knee Angular Velocity	0.314 (0.169 - 0.511)	0.892 (0.734 - 0.959)	0.655 (0.273 - 0.856)	0.839 (0.719 - 0.995)	0.374 (0.253- 0.657)	0.674 (0.582 - 0.857)	0.724 (0.390 - 0.888)	0.769 (0.305 - 0.964)	0.768 (0.471 - 0.907)	0.833 (0.507 - 0.992)
	Peak Bicep femoris muscle tendon unit length	0.752 (0.441 - 0.900)	0.765 (0.522 - 0.987)	0.744 (0.427 - 0.897)	0.744 (0.427 - 0.897)	0.745 (0.428 - 0.897)	0.740 (0.422 - 0.901)	0.748 (0.434 - 0.899)	0.777 (0.610 - 0.989)	0.746 (0.430 - 0.898)	0.746 (0.430 - 0.898)
	Take off (% gait)	0.980 (0.946 - 0.993)	0.984 (0.956 - 0.994)	0.969 (0.917 - 0.988)	0.944 (0.853 - 0.979)	0.987 (0.964 - 0.995)	0.985 (0.960 - 0.994)	0.974 (0.932 - 0.990)	0.977 (0.937 - 0.991)	0.958 (0.889 - 0.984)	0.979 (0.944 - 0.992)

Appendices Three – Running kinematics with increasing velocity

There was an exponential increase in peak hip extension and knee angular velocity kinematics with increasing running speed between high- and low-risk groups.

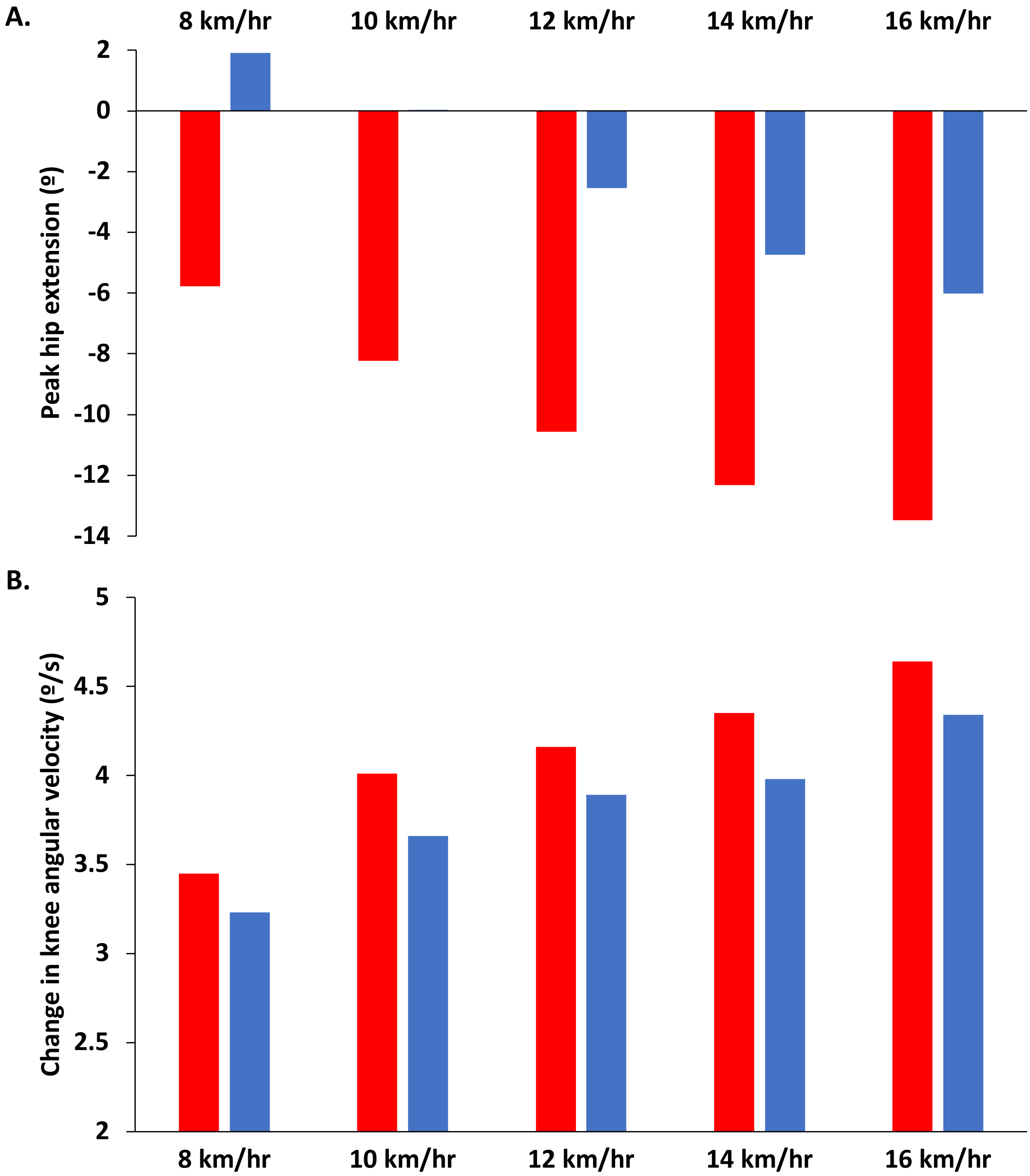

Figure Appendices three 1 - Increasing hip extension (A) and change in knee angular velocity (B) for high and low risk groups with increasing running velocity.

Furthermore, there were very large and nearly perfect positive relationships observed with increasing running velocity and magnitude of differences between the two risk groups. It would be expected that with further increases in running velocity up to sprinting, we may see further increases in the difference between high- and low-risk groups for kinematics presented here.

C.

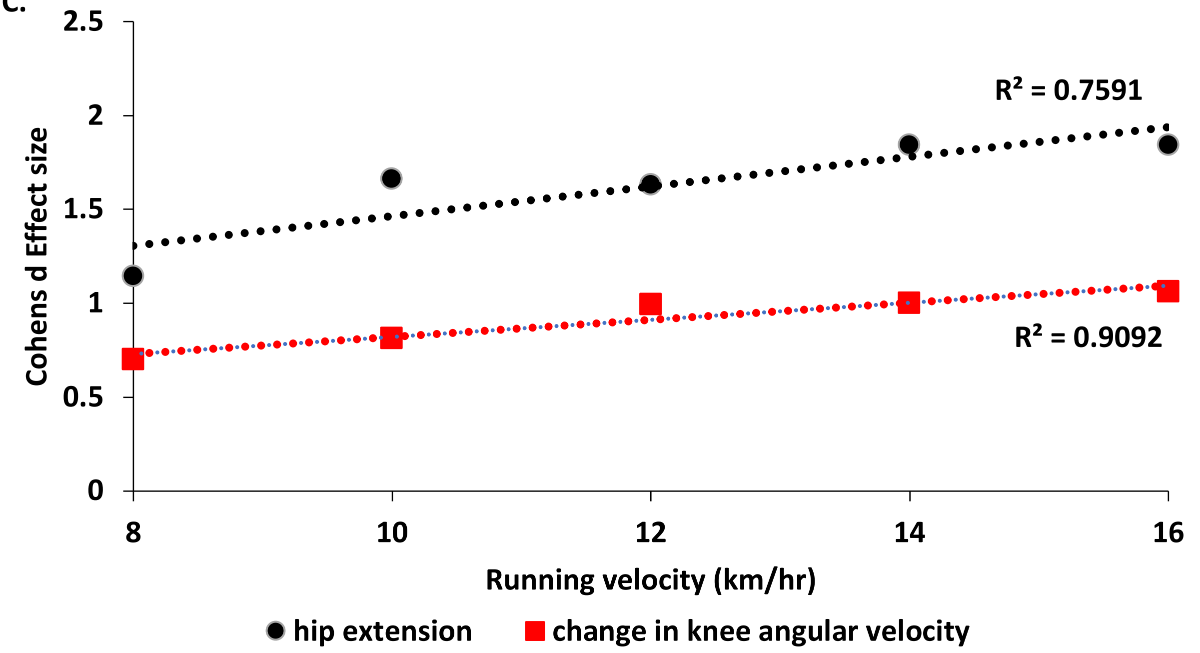

Figure Appendices three 2 - Very large (76% variance) and nearly perfect (91% variance) associations observed between the magnitude of difference between high- and low risk groups for peak hip extension and change in knee angular velocity with increasing running velocity.

Appendices Four – Changes in temporal characteristics from Pre- to Post-training (Chapter 7)

Individuals display varying strategies to performing all athletic tasks, and the NHE is no different. The training intervention performed with chapter 7, found significant increases in the active impulse and time to peak force for both intervention groups, with non-significant increases found for active impulse and time to peak force within the control group. However, all observed changes were small in magnitude (Hedge's *g* = 0.54 – 0.77) (Table 1, Figure 1 & 2), with moderate magnitude increase observed for the control group within active impulse.

Table Appendix Four Descriptive and statistical differences between time periods (PRE- and POST) for the NHE, Sprint and control training groups.

	Active Impulse ($N{\cdot}s^{-1}$)			Time to peak force (s)		
	NHE	**Sprint**	**Control**	**NHE**	**Sprint**	**Control**
Mean Difference	775.20	751.34	623.50	3.16	2.72	2.65
% Change	55.80	52.80	66.50	38.1	29.5	43.4
Hedge's *g*	0.60	0.68	0.89	0.61	0.54	0.77
p	<0.001	0.004	0.097	0.003	0.036	0.033

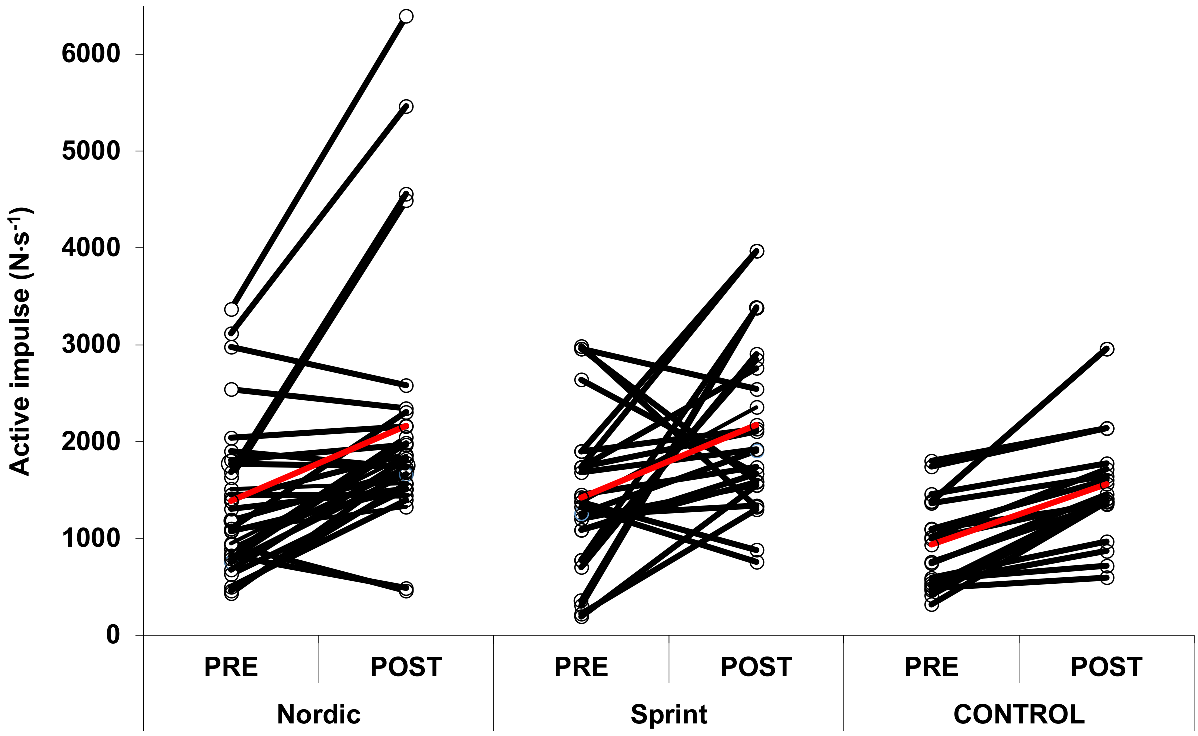

Figure Appendices four 1 - PRE and POST changes for the NHE, Sprint and Control training groups for active impulse.

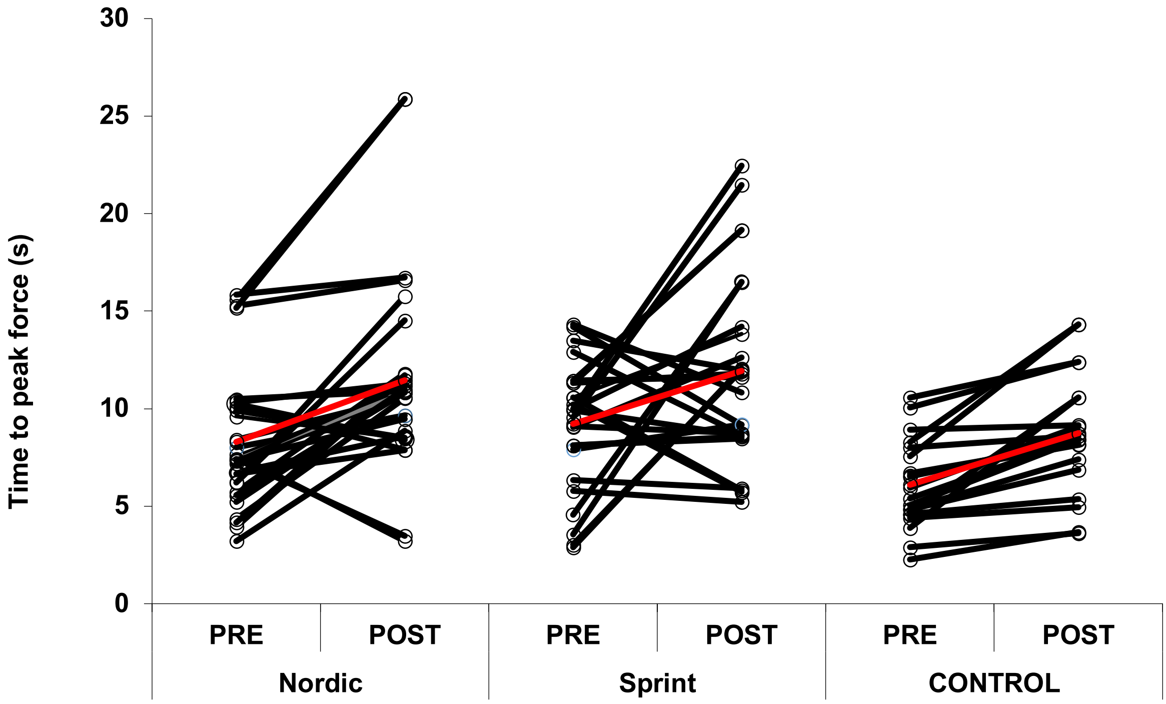

Figure Appendices four 2 - PRE and POST changes for the NHE, Sprint and Control training groups for time to peak force.

Appendices Five – Health Questionnaire

Health Questionnaire and Informed Consent

Participants Health Questionnaire

Surname : …………………………… Forename(s) : …………………………....
Date of birth : ………….................. Age : ………………………….…………..
Height (cm) : ……………………..… Weight (kg) : …………….………………….

2. Additional information

a. Please state when you last had something to eat / drink……………………...
b. circle the statement that relates to your present level of activity:
Inactive moderately active highly active
c. Give an example of a typical weeks exercise:
……………………………………………………………………………………...
d. If you smoke, approximately how many cigarettes do you smoke a day……..

3.	Are you currently taking any medication that might affect your ability to participate in the test as outlined?	YES	NO
4.	Do you suffer, or have you ever suffered from, cardiovascular disorders? e.g. Chest pain, heart trouble, cholesterol etc.	YES	NO
5.	Do you suffer, or have you ever suffered from, high/low blood pressure?	YES	NO
6.	Has your doctor said that you have a condition and that you should only do physical activity recommended by a doctor?	YES	NO
7.	Have you had a cold or feverish illness in the last 2 weeks?	YES	NO

8.	Do you ever lose balance because of dizziness, or do you ever lose consciousness?	YES	NO

9.	Do you suffer, or have you ever suffered from, respiratory disorders? e.g. Asthma, bronchitis etc.	YES	NO
10.	Are you currently receiving advice from a medical advisor i.e. GP or Physiotherapist **not** to participate in physical activity because of back pain or any musculoskeletal (muscle, joint or bone) problems?	YES	NO
11.	Do you suffer, or have you ever suffered from diabetes?	YES	NO
12.	Do you suffer, or have you ever suffered from epilepsy/seizures?	YES	NO
13.	Do you know of any reason, not mentioned above, why you should not exercise? e.g. Head injury (within 12 months), pregnant or new mother, hangover, eye injury or anything else.	YES	NO

EFFECT OF FILTERING WINDOW DURATIONS ON PEAK AND MEAN ELECTROMYOGRAPHY AMPLITUDE OF THE BICEP FEMORIS DURING THE GLUTE-HAM RAISE EXERCISE.

Journal of Strength and Conditioning Research: February 2019 - Volume 33 - Issue 2 - p e3-e217 doi: 10.1519/JSC.0000000000002990

N.J. Ripley, M. Cuthbert, N. Walker, J.J. McMahon, P. Comfort.

Introduction: Numerous investigators have used root mean square (RMS), as a method of filtering raw data from electromyography (EMG). However, for assessment of muscular activity, during dynamic tasks, there has been no standardization regarding the duration of the moving average window (MAW) used, ranging from 20 –200 ms. **Purpose:** To determine the effect of using different MAW durations, when filtering raw EMG data. **Methods:** Resistance trained individuals (n = 13, age: 23 ± 4 years; mass: 75.15 ± 9.65 kg; height: 1.76 ± 0.07 m) participated in this study by performing three repetitions of the glute-ham raise. Following standardized skin preparation, Ag-AgCl electrodes and wireless EMG sensors were attached to the bicep femoris, parallel with the orientation of the muscle fibers and in a bipolar configuration, with an inter-electrode distance of 17.5 mm, in accordance with SENIAM guidelines**.** Raw EMG data was captured at 1500 Hz, with high- and low-pass filtering between 10 and 1000 Hz. RMS values were calculated in a custom Excel spreadsheet, using MAW durations of 25, 50, 100, 200 and 400 ms. Means and standard deviations (SD) were determined for each MAW duration for peak and mean EMG amplitudes. Within-session reliability was assessed via intraclass correlation coefficients (ICC) and coefficient of variation (%CV). Minimum acceptable reliability was determined with an ICC ≥0.8 and CV <10%. Standardized differences were calculated using Cohen's *d* effect sizes, interpreted as trivial <0.19, small 0.20–0.59, moderate 0.60–1.19, large 1.20–1.99, very large >2.0. Multiple one way repeated measures analysis of variance, with Bonferroni post hoc analyses, were conducted to determine differences in EMG amplitudes values between MAW durations. An *a prior* alpha level was set at $p \leq 0.05$. **Results:** The results of this study demonstrate that all MAW durations result in highly reliable measures for both peak and mean EMG amplitudes, with low variability (table 1). Peak and mean EMG amplitudes were significantly different between all MAW durations, with the greatest differences found between 25 vs 400, across all EMG measures. **Conclusions:** As different MAW durations result in significantly different EMG amplitudes, with the greatest peak amplitude occurring with a MAW duration of 25 ms and lowest at 400 ms; this demonstrates that comparisons between the data from previous studies should be made with caution. The results of the current study are in contrasts to previous literature, identifying those high levels of reliability and low levels of variability can still be achieved with small MAW durations. **Practical Applications:** When assessing the muscle activation during resistance-based exercises, all MAW durations could be used reliably. However, the smallest MAW duration of 25 ms should be standardized as it demonstrates the greatest reliability, it would also allow for accurate identification of different phases of movement with a smaller smoothing effect.

Table. Mean, standard deviation (SD), within-session reliability and pairwise comparisons across filtering windows for EMG amplitudes of the Bicep Femoris during the Glute-Ham Raise exercise

	Peak EMG					**Mean EMG**				
Window (ms)	**25**	**50**	**100**	**200**	**400**	**25**	**50**	**100**	**200**	**400**
Trial 1	677.21	558.82	485.3	426.55	382.87	77.92	80.1	81.47	82.33	79.01
Trial 2	644.61	522.06	456.42	403.62	368.51	76.43	78.89	80.26	80.86	78.28
Trial 2	662.61	548.65	474.56	420.64	379.5	71.92	74.32	75.61	75.72	73.21
Mean (mv)	661.48	543.18	472.09	416.94	376.96	75.42	77.77	79.11	79.64	76.83
SD	16.33	18.99	14.6	11.91	7.51	3.12	3.04	3.09	3.47	3.16
%CV	2.47	3.5	3.09	3.86	1.99	4.14	3.91	3.91	4.36	4.11
ICCs (95% CI)	0.883 (0.790 - 0.941)	0.850 (0.737-0.924)	0.848 (0.733-0.923)	0.869 (0.767-0.934)	0.877 (0.781 - 0.938)	0.889 (0.801 - 0.945)	0.897 (0.814 - 0.949)	0.917 (0.848 - 0.959)	0.894 (0.809 - 0.947)	0.897 (0.814 - 0.948)
	p			***d***		***p***			***d***	
25 vs 50	0.004			6.68		0.014			0.76	
25 vs 100	0.012			12.23		0.005			1.19	
25 vs 200	0.002			17.11		0.024			1.28	
25 vs 400	0.003			22.39		0.256*			0.45	
50 vs 100	0.015			4.20		0.040			0.64	
50 vs 200	0.010			7.97		0.171*			0.57	
50 vs 400	0.016			11.51		0.298*			0.45	
100 vs 200	0.011			4.14		1*			0.16	
100 vs 400	0.019			8.20		0.044			0.73	
200 vs 400	0.040			4.02		0.084*			0.85	

SD = standard deviation; CV = coefficient of variation; ICC = intraclass correlation coefficient; CI = confidence interval; *d* = Cohen's *d* effect size; * no significant difference

EFFECT OF DIFFERENT ONSET THRESHOLDS ON ELECTROMYOGRAPHY VARIABLES OF THE BICEP FEMORIS DURING THE GLUTE-HAM RAISE EXERCISE.

Journal of Strength and Conditioning Research: February 2019 - Volume 33 - Issue 2 - p e3-e217 doi: 10.1519/JSC.0000000000002990

N.J. Ripley, J.J. McMahon, N. Walker, M. Cuthbert, P. Comfort.

Introduction: Electromyography (EMG) has been regularly used to assess muscular activity during dynamic tasks, including resistance exercises. One problem that currently compromises such research is how the onset of activation is identified, with no consistency across studies. **Purpose:** To examine the effect of using different onset thresholds on EMG variables. **Methods:** Resistance trained individuals (n = 13, age: 23 ± 4 years; mass: 75.15 ± 9.65 kg; height: 1.76 ± 0.07 m) participated in this study. Following a standardized skin preparation, Ag-AgCl electrodes and wireless EMG sensors were attached to the bicep femoris, in accordance with SENIAM guidelines; attached parallel to the orientation of the muscle fibers, in a bipolar configuration, with an inter-electrode distance of 17.5 mm**.** Raw EMG data were captured at 1500 Hz, with high- and low-pass filtering between 10 and 1000 Hz. Onset thresholds were calculated in a custom Excel spreadsheet, using calculations of; standard deviation (SD) of a resting baseline plus the mean baseline EMG (1-, 2- and 3 x SD + mean), mean baseline EMG plus an arbitrary value (mean + 0.015 mv), and percentage (10%) of the peak EMG during the task. Mean and SD were determined for peak EMG amplitude, mean EMG task amplitude and time of activation onset. Within-session reliability was assessed via intraclass correlation coefficients (ICC) and coefficient of variation (CV). Acceptable reliability was determined with an ICC ≥0.8 and CV <10%. Standardized differences were calculated using Cohen's *d* effect sizes. Multiple one-way repeated measures analysis of variance with Bonferroni post hoc analyses, were used to determine differences in EMG variables between different onset thresholds. An *a priori* alpha level was set at $p \leq 0.05$. **Results:** Different onset thresholds had no effect on both initial peak and mean EMG amplitudes, therefore were not taken forward for further analysis (table 1). Pair wise comparisons between 1 x SD + mean baseline vs 10% peak task and mean baseline + 0.015 mv vs 10% peak task, identified a significant delay in the time to activation when using 10% peak task. All other pairwise comparisons showed no significant difference. **Conclusions:** High levels of reliability were found when measuring EMG amplitudes. The highest reliability for time of activation was found for mean baseline + 0.015 mv and 10% of peak task, this is understandable as the SD is not taken into account. Different onset thresholds resulted in no difference between EMG amplitudes and no significant differences between time of activation for all but two pairwise comparisons. **Practical Applications:** The onset threshold used has no effect on task EMG amplitudes; however, they did affect the time of activation. If the time of activation is an important variable it is advisable to use an onset threshold calculation that produces the greatest reliability, e.g., mean baseline + 0.015 mv or 10% of peak task.

Table. Mean, standard Deviation (SD), within-session reliability and pairwise comparisons for EMG variables of the Bicep Femoris during the Glute-Ham raise exercise using different onset threshold calculations

	T1	T2	T3	Mean	SD	%CV	ICC (95% CI)
Peak (mv)	236.38	253.03	248.76	246.06	8.64	3.51	0.882 (0.788 0.941)
Mean (mv)	79.01	78.28	73.21	76.83	3.16	4.11	0.897 (0.814 - 0.948)
1 x SD + mean baseline (s)	2.76	2.65	2.46	2.63	0.15	5.81	0.674 (0.374-0.843)
2 x SD + mean baseline (s)	2.88	2.88	2.55	2.77	0.19	6.84	0.595 (0.377 - 0.773)
3 x SD + mean baseline (s)	2.96	2.85	2.58	2.80	0.20	7.02	0.556 (0.330 - 0.748)
Mean baseline + 0.015 mv (s)	2.45	2.46	2.32	2.41	0.08	3.36	0.741 (0.462-0.864)
10% Peak Task (s)	3.28	3.23	3.05	3.19	0.12	3.75	0.740 (0.501-0.875)
	p				***d***		
1 x SD + mean vs. 2 x SD + mean	0.749				0.82		
1 x SD + mean vs. 3 x SD + mean	0.229				0.96		
1 x SD + mean vs. mean + 0.015 mv	0.516				1.83		
1 x SD + mean vs. 10% peak task	0.015*				4.12		
2 x SD + mean vs. 3 x SD + mean	1				0.15		
2 x SD + mean vs. mean + 0.015 mv	0.311				2.47		
2 x SD + mean vs. 10% peak task	0.110				2.64		
3 x SD + mean vs. mean + 0.015 mv	0.331				2.56		
3 x SD + mean vs. 10% peak task	0.123				2.36		
Mean + 0.015 mv vs. 10% peak task	0.014*				7.65		

SD = standard deviation; CV = coefficient of variation; ICC = intraclass correlation coefficient; CI = confidence interval; *d* = Cohen's d effect size; * significant difference

A Systematic Review of Surface Electromyography Onset Activation Analysis Techniques During Running Tasks

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British Association of Sport and Exercise Scientists (BASES) Biomechanics interest group meeting 2017. University of Salford.

INTRODUCTION

Surface electromyography (sEMG) is often used as a method of assessing relative muscle activation in a range of athletic and occupational tasks (Ball & Scurr, 2011). This process facilitates understanding of the neuromuscular requirements of the assessed task and can aid in identifying relative muscular contributions during selected activities (Ball & Scurr, 2011).

Temporal characteristic analysis of sEMG data, allows for the identification of the onset of activation as well as the ability to identify muscle activation patterns which may occur during specific tasks (Ball & Scurr, 2011). This requires an accurate identification of the onset of muscle activation, via the use of a specific amplitude onset threshold. However, within the literature, a number of different calculations have been used to identify muscle activation onset thresholds. Therefore, the aim of this review was to explore the methods that have been used within the literature to calculate muscle activation onset thresholds during running.

METHODS

We searched for "EMG onset threshold", "muscle activation" and "running" using popular databases (Google Scholar, PubMed and EBSCO). An exclusion criterion was used, where all articles were required to use surface EMG as a measure of muscle activation and to have identified the onset threshold calculation used. A total of 454 non-duplicate journal articles were identified, with 447 excluded through screening. Which resulted in the inclusion of seven journal articles in this review.

RESULTS

After completion of the review process only seven methods have been reported to determine the onset of muscle activation (Table 1).

DISCUSSION

Across the seven methods reported within the literature, the majority used a measure of a resting baseline EMG amplitude to ascertain a specific threshold value of muscle activation onset. Two further methods included the use of visual inspection of activation onset and a percentage of peak task activation onset threshold. The use of visual inspection to determine activation onset results in construct validity issues, due to the inability of the investigators to be able to identify the specific time point of muscle activation. Furthermore, the use of an arbitrary percentage value of the peak task EMG, may also impact on the construct validity, as an individual's baseline resting EMG could potentially be greater than the set percentage value resulting in a false positive result.

CONCLUSION

There is no consensus regarding the criterion method of activation onset threshold of those previously reported within the literature. Furthermore, the method used for onset threshold can influence activation onset detection (Winter, 1984), and subsequently the accurate measurement of task activation onset. Therefore, future research should attempt to identify an optimal method of standardising the identification of an activation onset threshold.

REFERENCES

Ball & Scur, (2011). *Eur J Sp Sc.*11(6):447-56.
Winter, (1984). Arch Phys Med Rehabil. 65(7):393-8.

Table. Identification of the different onset threshold methods used within the literature

Study	Onset Method	Muscles analysed
McKinlen & Pedotti, 1992	> upper 95% CI for baseline for more than 10 ms	BF, VL, RF, LG, SOL, TA
Nyland et al., 1994	> 3 x SD over mean baseline	RF, VL, VM, MH, MG
Anderson, Nillsson & Thorstensson, 1997	Visual inspection (two investigators)	IL, PS, SA, RF, TF
Neptune, Wright & Van Den Bogert, 1998	> 3 x SD over mean baseline for longer than 50 ms	SOL, MG, TA, PL, VM, VL, RF, Gmax, Gmed
Kato & Ohtsuki, 2000	Visual inspection (one investigator)	VM, LG, Gmed, SAR
O'Connor & Hamill, 2004	> 10% Peak Task EMG	TA, PL, LG, MG, SOL
Karamanidis, Arampatziz & Bruggemann, 2004	> (mean baseline + (2 x SD)) over mean baseline	TA, LG, Gmed, VL, H

CI = confidence interval, SD = standard deviation, ms = milliseconds, Gmax = Gluteus maximus, Gmed = Gluteus medius, BF = Bicep femoris, MH = medial hamstrings, H = hamstrings, VM = vastus medialis, VL = vastus lateralis, RF = rectus femoris, PS = Psoas, IL = Illiacus, SAR = Sartorius, MG = medial gastrocnemius, LG = Lateral Gastrocnemius, SOL = Soleus, TA = tibialis anterior, PL = peroneus longus

EFFECT OF THE NORDIC HAMSTRING EXERCISE ABILITY ON IN-VIVO FASCICLE DYNAMICS DURING VARIATIONS OF THE NORDIC HAMSTRING EXERCISE

Ripley, Nicholas J.; Comfort, Paul; and McMahon, John J. (2020) "EFFECT OF THE NORDIC HAMSTRING EXERCISE ABILITY ON IN-VIVO FASCICLE DYNAMICS DURING VARIATIONS OF THE NORDIC HAMSTRING EXERCISE.," ISBS Proceedings Archive: Vol. 38 : Iss. 1 , Article 103.

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The purpose of this study was to determine if the ability to perform the Nordic hamstring exercise (NHE) impacts upon the fascicle dynamics of the bicep femoris long head during the NHE performed flat, decline and incline angles. 10 physically active individuals (8 males and 2 females, age 24.1±3.9 years, body mass 81.8±8.9kg, height 178.8±7.7cm) with a history of performing the NHE for training, were separated into two equal groups of high and low performers of the NHE via break-point angle assessed using 3D motion capture. Dynamic ultrasound (US) videos were collected using a 10cm probe, while semi-automatic software was used to analysed the fascicle changes. Fascicle lengthening during the NHE is dependent on NHE performance ability, with likely differences (non-overlapping control limits) between high and low performers. While absolute fascicle change was greater in the incline NHE for low performers, greater FL change was observed in the flat NHE for high performers. This could be as a result of the high performers possessing greater resting fascicle length and eccentric hamstring strength.

KEYWORDS: Hamstrings, dynamic ultrasound, bicep femoris, fascicle length tracking.

INTRODUCTION: The plasticity of the hamstring muscles' fascicle length (FL), in response to different training stimuli, is extremely important in the reduction of HSI risk (Bourne et al., 2018). Supramaximal eccentric exercises, i.e. Nordic hamstring exercise (NHE), have been shown to increase bicep femoris (BF) FL due to the addition of sarcomeres in-series (Bourne et al., 2018). Fascicle dynamics of eccentric hamstring exercises, utilising dynamic ultrasound (US), has only been reported in one study (Cataneo, 2018). However, the examined exercises were submaximal, performed at a low load and with minimal to no negative work. Despite this, the greatest fascicle lengthening occurred within the glider, followed by the diver and extender exercises (Cataneo, 2018). Unfortunately, due to poor video quality the authors could only observe differences in FL between images captured at the beginning and end of each exercise, thus the results should be interpreted with caution, as they do not represent fascicle behaviour throughout entire repetitions.

The NHE is generally prescribed on a flat horizontal surface, where the eccentric portion is performed with a controlled descent until a break point is reached. However, increases in eccentric hamstring strength have been shown to be related to an increased break point angle (Delahunt, McGroarty, De Vito, & Ditroilo, 2016), therefore, once athletes are able to perform most of the movement with control, there should be a progressive application of external load (Bourne et al., 2018). However, performing the NHE at an incline or decline allows for manipulation of the lever arm through which the centre of mass (with respect to the knee) is acting, thereby increasing or decreasing the amount of force required to control the descent of the centre of mass for any given knee angular displacement. A decline position would result in a greater load at a shorter muscle length, whereas an incline position would reduce the load and potentially result in a longer muscle length at any given angular displacement. However, to date, no research has attempted to quantify the BF FL changes during the NHE. Therefore, the aim of this study was to quantify BF FL changes during the NHE and determine if the ability to perform the NHE, defined by NHE break point angle, impacts FL changes. This information may aid researchers and practitioners in explaining why preferential adaptations (i.e., increased BF FL), may occur when utilising the NHE.

METHODS: Ten physically active individuals (8 males and 2 females, age 24.1 ± 3.9 years, body mass 81.8 ± 8.9 kg, height 178.8 ± 7.7 cm) with no history of lower-limb injury participated. All participants reported being physically active having a training history of performing the NHE. The study was approved by the institutional Ethics committee and conformed to the principles of the Declaration of Helsinki (1983). Prior to performing the NHE, a standardised warm up was performed consisting of two sets of ten repetitions of body weight squats, lunges and leg swings. To perform the NHE, participants were knelt on a padded bench (Power lift, Jefferson, IA, USA), with the ankles secured immediately superior to the lateral malleolus by ankle pads. Participants performed three repetitions of the NHE at each position (Nordic hamstring bench angle flat (0°), incline 20° and decline -20°), in a random order, with one-minute rest provided between each repetition and 2-3 minutes between each position.

Three-dimensional lower limb motion data were acquired for the NHE variations via infrared Oqus cameras (Qualisys, Partille, Sweden) and Qualisys C-motion software (version 3.90.21, Gothenburg, Sweden). Passive retro-reflective markers were placed upon the lateral malleoli, lateral femoral epicondyles, greater trochanter and acromion process. Motion data were captured for 15 seconds, sampling at 250Hz. A linear array probe (10cm, 44Hz, Mylab 70 XVision, Genoa, Italy) collected dynamic US video clips from the participants' self-identified dominant leg. A custom designed cast was used to attach the probe to the posterior thigh ensuring adequate pressure. The probe was applied in orientation to the BF fascicles following the line of the muscle to enable optimal imaging through the entire movement. An external synch pulse was applied to both the US scanner along with an open analogue channel into the Qualisys software (error <0.002s). This synch pulse provided a matched time whereby the US images could be synchronised to the 3D motion for appropriate analysis and interpretation. Instantaneous hip and knee angles and knee angular velocity were calculated. Raw data was subsequently exported into a custom designed Excel spreadsheet, where movement onset was identified when participants moved >5° from a knee angle taken from the first two-seconds of data collection. To identify the instance where participants could no longer control the decent (i.e. break-point), a knee angular velocity threshold of $20°.s^{-1}$ was applied (Delahunt et al., 2016). Dynamic US videos were analysed using a semi-automated tracking algorithm (Ultratrack, MATLAB, Math-works) (Farris & Lichtwark, 2016). Video files were initially cropped corresponding to the points between movement onset and break point. Following this, a muscle region of interest and fascicle end points were defined. The muscle region of interest was defined as the area between the superficial and deep aponeuroses of the BF. A muscle fascicle of interest was defined as the straight-line distance between the superficial and deep aponeuroses. A fascicle was chosen based on it being visible across the entire task for all participants.

Data are presented as mean ± SD. Normality was assessed by Shapiro-Wilk's statistic. Absolute and relative between-trial reliability were assessed by coefficient of variation (CV) percentages and a two-way random effect model intraclass correlation coefficient (ICC) with 95% CIs. Between trial reliability of time-series data from the US was assessed using a coefficient of multiple correlation (CMC) with 95% CIs. Minimum acceptable reliability was confirmed using an CV <10%. The ICC and CMC values will be interpreted based on the lower bound CI as (<0.50) poor, (0.5-0.74) moderate, (0.75-0.90) good and (>0.90) excellent (Koo & Li, 2016). Mean time-series data of high and low NHE performers change in FL was plotted along with the corresponding upper and lower 95% confidence intervals to create upper and lower control limits, where a likely difference is determined by non-overlapping shaded areas. The NHE performance was determined by break-point angle, with high (n=5) and low performers (n=5).

RESULTS: All data was normally distributed (*p* >0.05). Break point angle demonstrated high absolute and relative reliability (Table 1). The high performing group reached break point angles of 141 ± 4°, 129 ± 6° and 108 ± 11° for incline, flat and decline, respectively. In contrast however, the low performing group reached break point angles of 122 ± 13°, 103 ± 7° and 82± 13° for incline, flat and decline, respectively.

Table. Absolute and relative between-trial reliability for kinematic and dynamic ultrasound measures

	Break point angle		
	Incline	Flat	Decline
ICC (95%CI)	0.877 (0.626 - 0.975)	0.965 (0.877 - 0.993)	0.943 (0.809 - 0.989)
CV%	0.67	1.44	1.64
	Dynamic Fascicle change		
ICC (95% CI)	0.977 (0.965 - 0.989)	0.917 (0.816 - 1.000)	0.979 (0.963 - 0.995)
CMC (95% CI)	0.969 (0.958 - 0.981)	0.901 (0.802 - 1.000)	0.972 (0.958 - 0.985)

The performance groupings were identical across each position. Between trial time-series data for FL changes demonstrated nearly perfect relative reliability for all positions (Table1).

The higher performing groups (i.e., those who achieved a greater break-point angle) displayed greater FLs across all positions (Figure 1-3). The incline angle displayed likely differences across the entire normalised time-series (Figure 1), whereas both flat and decline variations displayed overlapping control limits, indicating non-likely differences within the time-series (Figures 2 & 3).

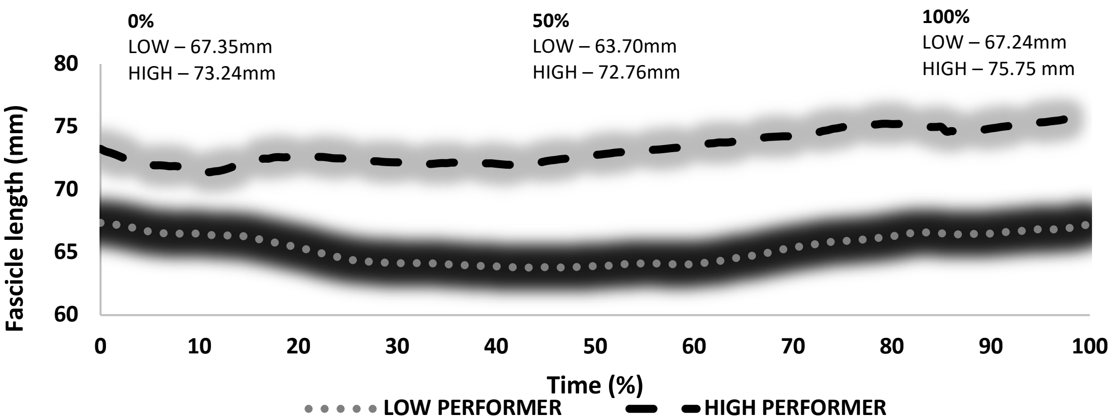

Figure 1. Dynamic BF FL changes during the incline NHE for high and low performers with 95% CIs.

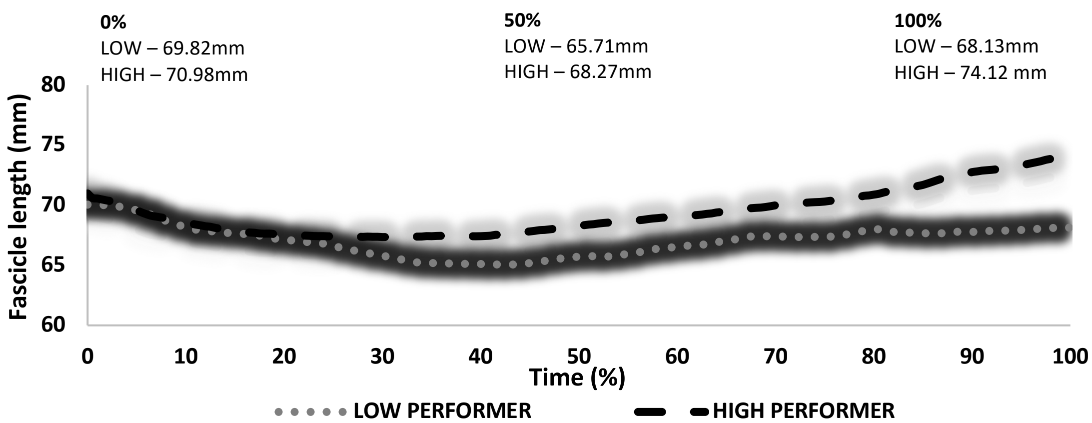

Figure 2. Dynamic BF FL changes during the flat NHE for high and low performers with 95% Cis.

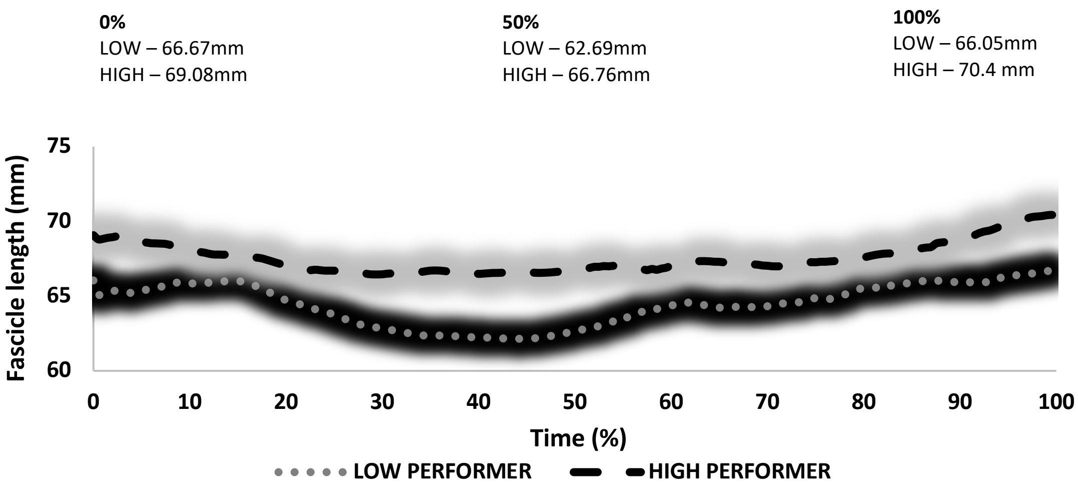

Figure 3. Dynamic BF FL changes during the decline NHE for high and low performers with 95% CIs.

DISCUSSION: The results of the present study demonstrate that an individual's NHE performance could alter the FL dynamics within the BF, as the higher performers possessed greater FLs and went through a greater degree of FL change throughout each of the NHE variations. In contrast, low performers who underwent similar initial shortening, only lengthened to their initial starting lengths up to the break point (100%) of the NHE. This finding could be explained by potential relationships between NHE break point angle, eccentric hamstring strength and BF FL. The high performers, who possessed the greater FLs, would also be expected to possess greater eccentric strength and capability to actively, control, lengthening prior to reaching a break point. The FLs observed within the present study are shorter than those previously reported for resting and 25% MVIC lengths (Bourne et al., 2018), however this is not surprising given the changes in anatomical position. The observed FLs are similar to those presented by Kellis (2018), who observed FL at different anatomical positions, during passive stretching.

This is the first study to investigate the effect of the performance angle on the NHE. For both groups, break-point angle was found to be greatest in the incline variations indicating that this variation could be of a lower intensity, permitting the participants to train at longer muscle lengths, a common complaint made by coaches about the NHE. Furthermore, during the incline variation the low performing group went through the greatest fascicle lengthening, albeit to return to the initial length. This finding is crucial, as controlled, lengthening is what is required for the desired adaptive response (increased FL and eccentric strength (Bourne et al., 2018)). Therefore, an incline variation that permits greater fascicle lengthening, under control could be more effective exercise for training, especially for lower performing individuals.

Dynamic ultrasound imaging is not a novel concept; however, this is the first study to analyse dynamic ultrasound videos of the BF during exercise, whereas previous attempts have only analysed single images (Cataneo, 2018). A number of methodological difficulties have been reported previously such as plane, depth, image quality and the fact that the FL often exceeds many ultrasound probes, requiring extrapolation increasing the potential sources of error. One explanation as to why this study succeeded with high levels of reliability, is that a 10 cm field of view which was utilised, was able to image the entire fascicle without the need for estimation equations, providing optimal image quality for the automatic processes.

CONCLUSION: Dynamic FL changes during the NHE could be dependent on NHE performance ability. Furthermore, alterations made to the position of the NHE can also impact upon FL changes, with the incline and flat variations permitting the greatest absolute FL change for the low and high performing groups, respectively. This of interest to practitioners as the desired adaptations, from the controlled lengthening action, could be optimised by altering the performance angle of the NHE, with an appropriate regression of intensity within the NHE being an incline for lower ability individuals.

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Retention of adaptations to eccentric hamstring strength and bicep femoris fascicle length from a seven-week training intervention including sprinting or Nordic hamstring exercise.

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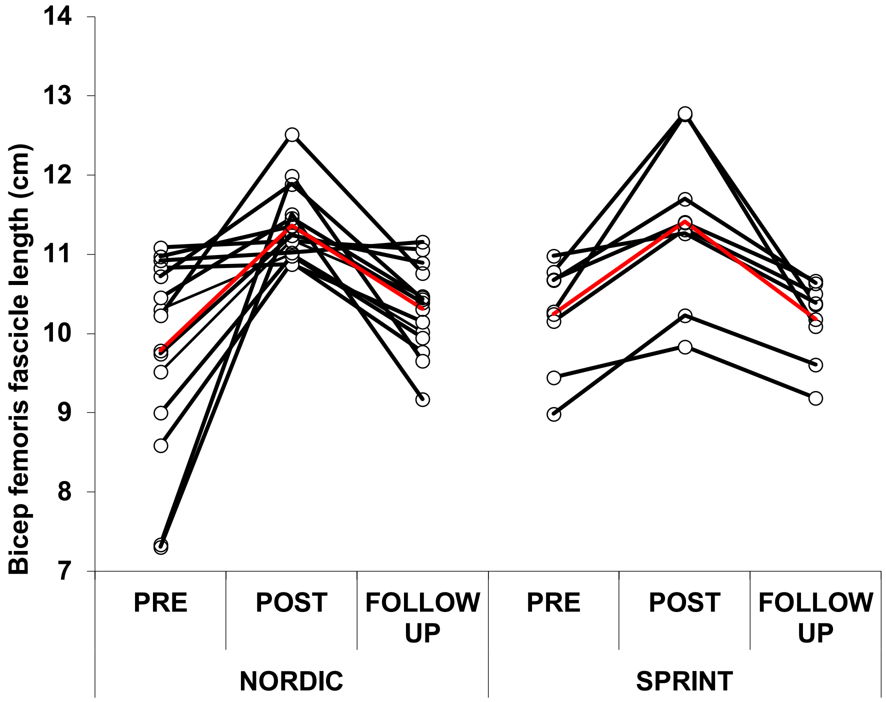
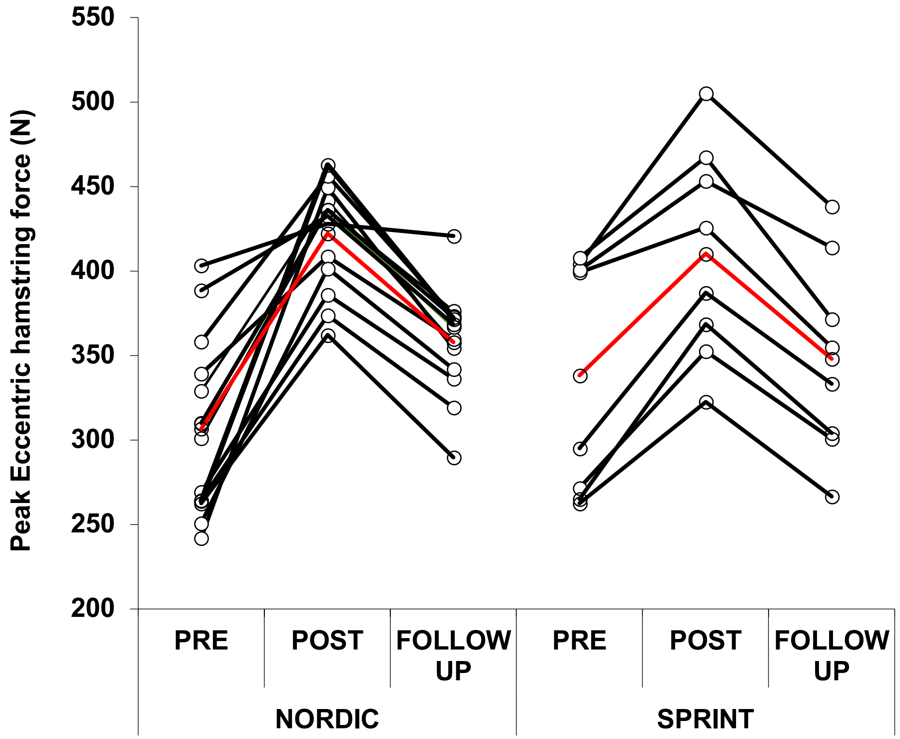
Eccentric hamstring strength (EHS) and bicep femoris fascicle length (BFL) have been shown to decrease within a two-week detraining period from the application of the Nordic hamstring exercise (NHE). However, alternative methods (i.e. sprinting (ST)) have not been explored with respect to detraining. **PURPOSE** This study aimed to observe the adaptation and retention of adaptations to EHS and BFL and sprint ability from either ST or NHE. **METHODS** 10 physically active individuals participated in this study and were randomly assigned into either NHE or ST groups. An identical resistance training program was performed twice per week, including clean derivatives, back squat, reverse lunge and Romanian deadlift, with the addition of the NHE or ST. Pre-, post-, and follow-up (FUP) testing included BFL was collected using a 10 cm ultrasound with images taken on the mid-point between ischial tuberosity and lateral epicondyle. Peak EHS was assessed by participants performing three repetitions of the NHE on the Nordbord, sampling at 50 Hz. BFL was analysed using ImageJ software and the following equation OFL+(h÷SIN(PA)), where OFL is the observed fascicle, h is the perpendicular distance between aponeurosis and BF end point and PA is the pennation angle. RMANOVAs were used to determine training induced changes in all tests. Post-hoc testing with Bonferroni corrections and Hedge's *g* effect sizes was performed to determine the magnitude of differences. An *a priori* alpha level was set at $p \leq 0.05$. Hedge's *g* Effect sizes interpreted as trivial (≤0.19), small (0.20–0.59), moderate (0.60–1.19) and large (>1.20). **RESULTS** A significant group x time interaction was found for peak EHS ($p=0.011$) (Figure 1). The NHE and ST groups had significant moderate-large increases in peak EHS ($p<0.001$, $g=0.77$-1.94), whereas at POST to FUP significant moderate-large decreases ($p<0.001$, $g= 0.74$-1.61), were observed. PRE to FUP a significant, moderate increase in peak EHS were observed for the NHE ($p<0.001$, $g=0.86$), while the ST groups had a non-significant trivial increase ($p=1.00$, $g=0.11$). A non-significant group x time interaction was found for BFL ($p>0.05$) (Figure 1). Significant, moderate-large increases in BFL were observed for the NHE ($p<0.001$, $g=1.12$) and ST groups ($p=0.020$, $g=1.01$). From POST to FUP significant, moderate-large decreases in BFL were observed for NHE ($p=0.003$, $g= 1.59$) and ST groups ($p=0.012$, $g= 0.98$). From PRE to FUP, a non-significant, moderate increase in BFL was observed for the NHE ($p=0.706$, $g=0.37$), while the ST groups had a non-significant, trivial decrease ($p=1.00$, $g=0.07$). **CONCLUSIONS** Both the NHE and ST can improve the modifiable risk factors of HSIs. However, decreases are seen after a two-week detraining period. **PRACTICAL APPLICATIONS** The NHE is more effective than ST, in retaining the adaptive response for both EHS and BFL, however, continual application of either intervention would be crucial to maintain EHS and BFL.



Bicep femoris fascicle length (cm)
14
13
12
11
10
9
8
7
PRE
POST
FOLLOW UP
PRE
POST
FOLLOW UP
NORDIC
SPRINT



Peak Eccentric hamstring force (N)
550
500
450
400
350
300
250
200
PRE
POST
FOLLOW UP
PRE
POST
FOLLOW UP
NORDIC
SPRINT

Eccentric hamstring strength and sagittal plane lower limb running kinematics across team sports.

Ripley, N.J., Comfort, P., McMahon, J.J.

UKSCA Conference Podium presentation, Online September 2020

Individuals with impaired hamstring functioning, through a history of injury or acute fatigue, demonstrate alterations in running kinematics. However, specific demands of team sports such as football, rugby and court-based team sports (CBTS) (e.g., futsal, basketball), could lead to differences in running kinematics. Furthermore, hamstring strength has been suggested to play an important role in running performance and contributes to pelvic control, which can have a large influence on running kinematics and potentially influence hamstring injury occurrence. Therefore, the purpose of this study was to determine differences in eccentric hamstring strength and lower limb running kinematics between sports.

Sixteen collegiate team sport athletes (rugby n=5, 23.80±2.95 years, 185.10±6.58 cm, 92.16±12.10 kg, football n=7, 23.14±2.91 years, 179.14±6.09 cm, 84.37±10.93 kg, CBTS n=4, 26.75±3.92 years, 179.50±4.93 cm, 90.75±6.45 kg), participated within the present study, attending the laboratory on two separate occasions. During occasion one, peak relative eccentric hamstring torque (strength) was assessed using an isokinetic dynamometer sampling at $60°\cdot s^{-1}$. During the second occasion, lower extremity, 3D motion data was collected while participants completed a 15-s running trial on a treadmill at 16 $km\cdot hr^{-1}$, within a calibrated area of 10 infrared cameras (250 Hz). Running gait from three strides was time normalized from 0-100%, from touch down to subsequent touch down, with mean sagittal hip and knee angle plotted for each sport. A one-way ANOVA with Bonferroni post-hoc and Hedge's *g* effect sizes were conducted to compare mean differences between eccentric hamstring strength and peak hip and knee angles.

CBTS athletes achieved a significantly lower peak hip flexion in comparison to football ($p<0.001$) and rugby ($p = 0.0015$), to very large magnitude (g=2.97-2.99). Alternatively, rugby athletes achieved a significantly greater peak knee flexion in comparison to football and CBTS ($p<0.001$), with very large magnitude (g=2.56-5.89). Non-significant ($p >0.05$), trivial-moderate differences (g=0.13-1.12), were observed for peak relative eccentric hamstring strength, and peak hip and knee extension between sports, in addition to peak hip flexion between rugby and football and peak knee flexion between football and CBTS.

Various team sports place specific movement demands upon athletes, specifically with regards to available space and the potential of contact situations. The results of the present study indicate that these demands have a significant influence on running kinematics, despite similar relative eccentric hamstring strength, in non-sporting situations. These identified differences could be contributing to differences in pelvic control, thus influencing hamstring injury occurrence.

Sprint vs Nordic hamstring exercise training effect on eccentric hamstring strength and bicep femoris fascicle length: Effect of initial sprint ability on the magnitude of adaptations.

Nicholas Ripley, Paul Comfort, John McMahon

ALTIS Virtual Apprentice Coach Program, Online September 2020.

Introduction

Eccentric hamstring strength (EHS) and bicep femoris fascicle length (BF_L) are known risk factors for hamstring strain injuries (HSI) in team sports (TS). The Nordic hamstring exercise (NHE) and sprint training (SPT) can have positive adaptations on EHS and BF_L, however, the effect of initial sprint ability has not been observed. The purpose of this study was to compare the effect of initial sprint ability, when performing either the NHE or SPT, on EHS and BF_L.

Methods

28 TS athletes performed a control lower-limb resistance program for 7-weeks (2/week), with either additional SPT (n=13,22.2±2.5yrs,1.7 ± 0.05m,70.6±7.8kg) included 4-7x25-30m or NHE (NHE n=15,21.4±2.6yrs,1.7±0.04m,76.9±14.2kg) included 2x4 repetitions, which was assigned randomly. Sprint times were recorded using timing cells (0-10m, 0-20m and 10-20m). EHS was determined as the peak force during the NHE. BF_L was assessed using a 10-cm ultrasound probe. Initial sprint ability groups were determined by the median values. Two-way analysis of variance, with post-hoc analysis and Hedge's *g* effect sizes were performed.

Results

Faster athletes achieved greater increases in EHS and BF_L for SPT, in comparison to NHE (Table 1). However, the NHE was more effective for the slower athletes, in comparison to SPT.

Discussion

The NHE and SPT are both effective at increasing EHS and BF_L, however initial sprint ability influences their effectiveness. SPT is more effective for faster athletes, whereas the NHE was more effective for slower athletes. This information could aid TS, by minimising in-season eccentric loading associated with soreness, while maximising adaptations to reduce HSI risk.

Table. Median sprint times and interquartile range for the sample and between group, descriptive and statistical differences between PRE and POST training intervention for Nordic hamstring exercise and sprint training groups, differentiated by initial sprint ability.

		0-10m				0-20m				10-20m			
Median Sprint time (Interquartile range)		1.98 (0.09)				3.39 (0.12)				1.40 (0.10)			
		FAST		**SLOW**		**FAST**		**SLOW**		**FAST**		**SLOW**	
		NHE (n=9)	**SPT (n=7)**	**NHE (n=6)**	**SPT (n=6)**	**NHE (n=8)**	**SPT (n=7)**	**NHE (n=7)**	**SPT (n=6)**	**NHE (n=9)**	**SPT (n=5)**	**NHE (n=6)**	**SPT (n=8)**
	Sprint time (s)	1.92 (0.09)	1.91 (0.07)	2.00 (0.10)	2.09 (0.06)	3.25 (0.13)	3.29 (0.11)	3.53 (0.13)	3.67 (0.13)	1.34 (0.05)	1.36 (0.04)	1.47 (0.09)	1.55 (0.08)
Eccentric hamstring strength	**PRE (N)**	338.76 (64.07)	318.97 (54.65)	286.12 (46.49)	268.76 (88.84)	327.7 (71.82)	322.1 (322.10)	306.28 (51.32)	265.12 (85.49)	319.08 (71.99)	271.25 (35.14)	315.65 (49.22)	311.14 (88.97)
	POST (N)	446.12 (59.86)	406.78 (39.30)	409.03 (58.85)	368.41 (74.14)	440.74 (69.40)	405.72 (38.97)	420.47 (51.12)	369.65 (75.09)	435.85 (66.56)	366.92 (28.90)	424.43 (54.81)	402.92 (69.88)
	p	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001
	Hedge's g (95% CI)	1.65 (0.29-2.79)	1.73 (0.56-2.94)	2.14 (0.94-3.26)	1.12 (-0.31-2.49)	1.51 (0.18-2.78)	1.63 (0.56-2.72)	2.09 (1.23-3.18)	1.20 (-0.19-2.50)	1.60 (0.22-2.78)	2.64 (1.12-4.36)	1.93 (1.03-2.89)	1.08 (0.07-2.07)
	Percent change (%)	33.84	29.19	43.55	34.44	36.78	27.70	38.81	36.18	39.21	36.34	35.50	28.66
Bicep femoris fascicle length	**PRE (cm)**	10.20 (0.89)	9.85 (0.48)	9.54 (0.63)	9.33 (0.67)	9.81 (1.33)	9.89 (0.62)	9.56 (0.57)	9.39 (0.72)	9.76 (1.26)	9.52 (0.63)	9.60 (0.62)	9.66 (0.54)
	POST (cm)	11.09 (0.86)	11.05 (0.92)	10.60 (0.49)	10.30 (0.60)	11.16 (0.95)	11.04 (0.93)	10.82 (0.42)	10.26 (0.50)	11.06 (0.94)	10.74 (0.73)	10.91 (0.39)	10.59 (0.99)
	p	0.003	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	0.004	<0.001	<0.001	0.008	0.028
	Hedge's g (95% CI)	0.97 (-0.12-1.79)	1.53 (0.72-2.24)	1.74 (0.44-3.10)	1.42 (-0.13-2.60)	1.10 (0.02-1.88)	1.37 (0.49-2.33)	2.36 (1.01-3.81)	1.29 (-0.01-2.69)	1.13 (0.16-1.91)	1.62 (0.48-2.76)	2.13 (0.96-3.36)	1.16 (0.17-2.20)
	Percent change (%)	8.98	12.11	10.97	10.50	9.73	11.60	11.82	9.40	9.81	12.81	12.05	9.56