The Association between Running Kinematics and Common Overuse injuries in Runners. Implications for Injury and Rehabilitation.

Mr Christopher Bramah

Doctor of Philosophy (PhD)

University of Salford

School of Health Sciences

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iv. Abbreviations

AKPS – Anterior knee pain scale

ANOVA – Analysis of Variance

AT – Achilles tendinopathy

BMI – Body mass index

CI – Confidence Interval

CPD – Contralateral pelvic drop

GPS – Global positioning system

GRF – Ground reaction force

HADD – Hip adduction

HIR – Hip internal rotation

IA – Initial assessment

ICC – Interclass Correlation Coefficient

ITB – Iliotibial band

ITBS – Iliotibial band syndrome

KMPW – Kilometres per week

LEFS – Lower extremity functional scale

MCID – Minimal clinically important difference

MD – Mean difference

MDC – Minimal Detectable Change

ML – mediolateral

MPW – miles per week

MR – Magnetic Resonance

MTSS – Medial tibial stress syndrome

NRS – Numerical rating scale

OR – Odds Ratio

PFJRF – Patellofemoral joint reaction force

PFP – Patellofemoral Pain

ROC – Receiver Operator Curve

SEM – Standard Error of measurement

SD – Standard Deviation

SS – Sum of squares

UK – United Kingdom

VAS – Visual analogue scale

v. Abstract

Background: Running related injuries are influenced by a complex interaction between multiple factors. Running kinematic patterns represent one such factor which will influence the load applied to musculoskeletal structures during each foot contact of a run. When combined with an increase in external training load, a cumulative tissue load may result that exceeds tissue capacity, resulting in injury development.

Aim: This thesis aimed to identify kinematic parameters associated with common running related injuries, explore whether such factors are influenced by training load exposure and investigate whether gait retraining, aimed at improving to running kinematics, may represent a clinically effective intervention.

Methods: A narrative literature review was conducted to identify gaps within the literature and formulate specific research questions. An initial study was performed to investigate the between day repeatability and quantify the standard error of measurement for discrete kinematic parameters during running. A case control study of 108 runners was then undertaken to investigate whether similar kinematic parameters are associated with multiple different common running related injuries. Following identification of kinematic parameters associated with running injuries, a cross sectional study investigated whether kinematic parameters associated with injury are associated with training load exposure. Finally, a case series study investigated whether gait retraining, in the form of a step rate intervention, improves running kinematics and clinical outcomes amongst a group of 12 injured runners with patellofemoral pain.

Findings: The repeatability study demonstrated good to excellent repeatability with low measurement errors for several kinematic parameters during treadmill running. The second study found several kinematic parameters to be associated with multiple different running related injuries, including increased contralateral pelvic drop, hip adduction and forward trunk lean, as well as reduced knee flexion and increased ankle dorsiflexion at initial contact. Within this study, a logistic regression analysis found peak contralateral pelvic drop to be the kinematic parameter most strongly associated with

common running injuries. Data from this study was used to set a critical threshold for peak contralateral pelvic drop, above which runners were deemed more likely to be at risk of injury. Building on this idea, the third study, highlighted an association between training load exposure and running kinematics. Specifically, amongst injury-free high-mileage runners, a significantly lower proportion of runners exhibited “high-risk” kinematics than in a group of injury-free low-mileage runners. Finally, the case series study found a single session of gait retraining, via a 10% increase in step rate, resulted in significant reductions in peak contralateral pelvic drop, hip adduction and knee flexion, as well as significant improvements in clinical and functional outcomes amongst runners with patellofemoral pain.

Implications: Several kinematic parameters appear to be associated with multiple different running related injuries, suggesting similar kinematic patterns may increase tissue load on multiple different anatomical locations. Interestingly, there appears to be a complex interaction between kinematics and training load exposure highlighting that kinematics alone may be unlikely to explain injury development. In such instances where runners have become injured and possess kinematic parameters which increase tissue load, increasing step rate appears to be an effective gait intervention which can be easily integrated into clinical practise and a runner’s normal routine.

Chapter 1: Introduction

1.1 Popularity and health benefits of running

Over the last decade running has become an increasingly popular method of physical activity. According to Sport England, more than 2 million people across the United Kingdom (UK) participate in running each week making running the most popular method of exercise amongst the UK population (2). Recreational running events are also increasing in popularity with a reported 347, 876 UK based runners applying for the London Marathon in 2019, a rise of over 20, 000 applicants compared to the previous year (3). The increasing popularity of running may in part be explained by the health and social benefits offered from this relatively inexpensive form of physical activity, including reducing body fat, lowering maximal heart rate, encouraging social interaction and benefits to mental health (4-6).

1.2 Injury Risk of Running

Despite numerous health benefits, running poses considerable risk of musculoskeletal injury. The overall incidence of running related injuries has been reported to range between 19.4 and 79.3% (7-10) with approximately 50% of runners injured annually and 25% being injured at any one time (11). In a recent retrospective study of 1145 UK based runners, 49.8% were reported to have a current injury, with 86% of injured runners continuing to run despite experiencing pain (8).

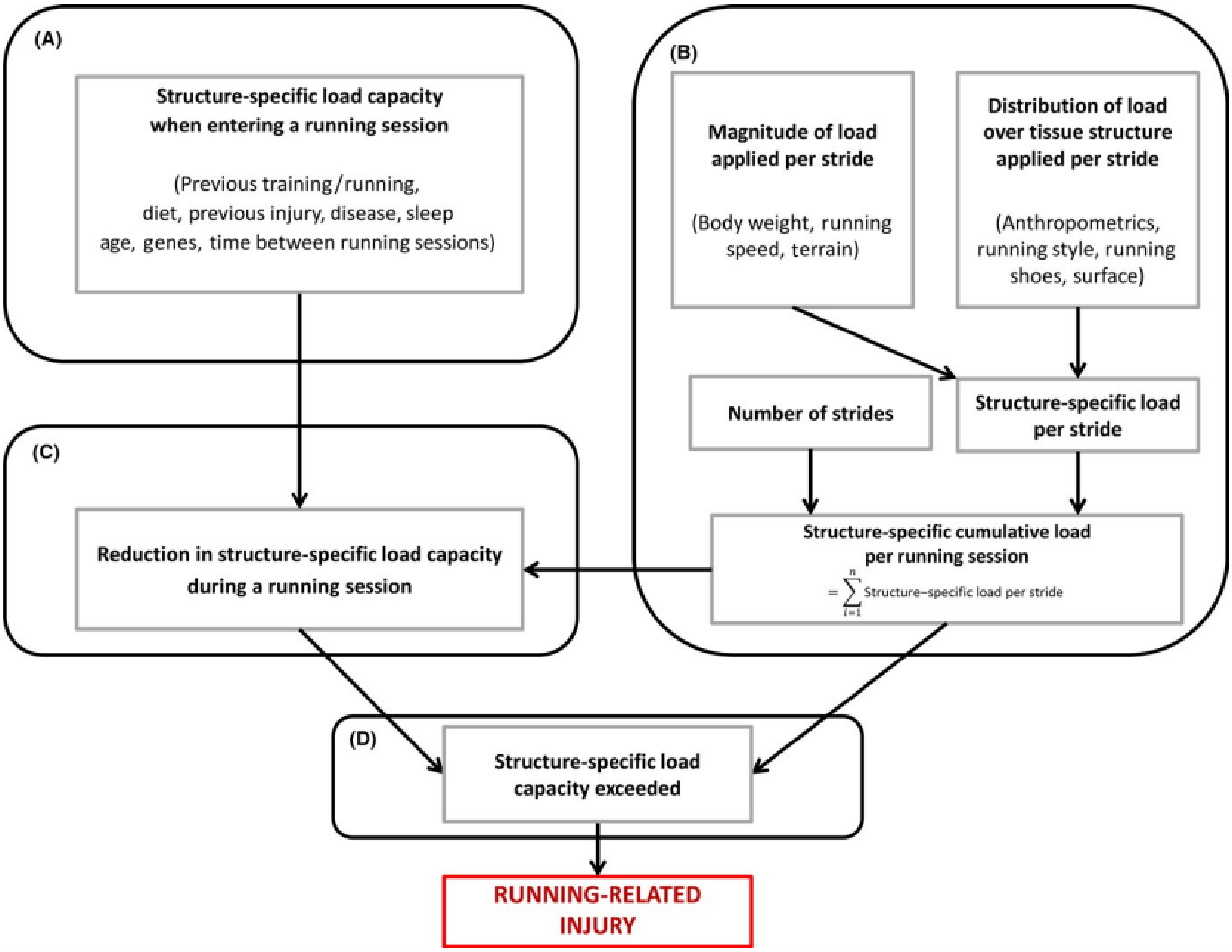
Of all running related injuries the majority are said to occur to the knee and lower limb, accounting for 7 to 50% and 9 to 32% of all injuries respectively (7). The most common injuries are cited as patellofemoral pain (PFP), iliotibial band syndrome (ITBS), medial tibial stress syndrome (MTSS) and Achilles tendinopathy (AT) (10, 12, 13). Incidence and prevalence rates have been reported to be as high as 20.8% and 22.7% for PFP (14), 9.1% and 12% for ITBS (12, 15), 20% and 9.5% for MTSS (12, 16) and 10.9% and 18.5% for AT (12). Many of these injuries are known to have lengthy recovery times (10), high reoccurrence rates and lead to a reduction or cessation of training in approximately 30 to 90% of cases (17). Furthermore, a running related injury has been reported to be the leading reason for discontinuing running, reducing the positive health effects of regular running (5).

1.3 Aetiology of Running Injuries

The majority of running injuries are considered overuse injuries with a multifactorial aetiology; influenced by the complex interaction between biological, psychological, behavioural and sociocultural factors (18, 19). Historically, many injury causation models focus predominantly on biomedical contributions to injury development (20, 21). One example being the biomechanical model of injury, where injury aetiology is viewed as the result of an imbalance between either the biomechanical loads applied to a tissue structure or the ability of the tissue to withstand the applied loads (1, 21, 22). Consequently, many studies focus on establishing causal connections between singular variables which may be associated with injury such as training errors (23), lower limb structure (24) or biomechanics (25, 26). Although this method of investigating injury is useful in establishing risk factors for injury, for which specific interventions can be targeted towards, it should also be interpreted amongst a wider context acknowledging the potential interaction between additional factors (19, 27).

Recently, Bertelsen et al (28) proposed a conceptual framework of running injury development, building upon the biomechanical model of injury to reflect the multifactorial and dynamic nature of running injury aetiology (Figure 1). Within this framework, the authors proposed that running related injuries occur due to the interaction between the structure specific load capacity of a musculoskeletal tissue, herein termed tissue capacity (Figure 1A), and the structure specific cumulative loads applied to a tissue during a run (Figure 1B), which will herein be referred to as the cumulative tissue load. If the cumulative tissue load exceeds tissue capacity (Figure 1D) then injury will occur. Importantly, the model highlights the multiple factors which influence both tissue capacity and cumulative tissue load, as well as the dynamic, adaptable and mal-adaptable, nature of tissue capacity. Understanding factors influencing both tissue capacity, cumulative tissue load and the interaction between factors, may aid understanding of injury causation and assist in the implementation of appropriately targeted rehabilitation interventions.

Figure 1 Conceptual framework for the aetiology of running related injuries reported by Bertelsen et al (28). Box A represents the structure specific tissue capacity at the beginning of a running session. Box B represents the factors influencing the structure specific cumulative load per running session. Box C represents the reduction in structure specific load capacity due to the interaction between box A and box B. Box D highlights how the interaction between structure specific load capacity and structure specific cumulative load may result in the structure specific load capacity being exceeded resulting in running related injury development.



1.3.1 Tissue Capacity

Tissue capacity is defined as the total load that a tissue can withstand before reaching its ultimate failure point (28-31). Importantly, tissue capacity is a dynamic construct changing both within a single loading bout and between subsequent loading bouts (Figure, 1A & 1C) (28, 32). This is largely dependent upon the tissue specific and individual response to load application, both acutely and over time (28, 31). According to Soligard et al (32) both physical and psychological response to load occurs along a continuum, progressing from tissue homeostasis to acute fatigue, functional and non-functional overreaching, overtraining syndrome, subclinical tissue damage and eventually clinical symptoms leading to a time loss injury or illness. Acutely, within a single running bout, repeated load application will result in a gradual reduction in tissue capacity, which could occur due to structural and/or mechanical changes to the tissue or individual, such as an increase in physiological fatigue impairing muscle function. Providing this loading bout is ceased before the cumulative load exceeds tissue capacity, and adequate recovery occurs between subsequent loading bouts, then tissue homeostasis is maintained, and tissue capacity may increase (28, 32). Conversely, if there is insufficient recovery between loading bouts, or the cumulative load with a single loading bout exceeds tissue capacity, tissue homeostasis is not maintained, tissue damage may occur, and tissue capacity will be gradually reduced. Consequently, musculoskeletal structures may become less tolerant to load resulting in the gradual progression of subclinical tissue damage, and with repeated loading bouts, the onset of injury (Figure 1D) (28, 32, 33).

Not only does tissue capacity vary between individuals, but it is also influenced by the interaction between a variety of biological, psychological and sociocultural factors (28-31). Several biological factors have been proposed to influence tissue capacity via their influence upon the mechanical and structural tissue properties and the tissue response to loading (31). These include non-modifiable factors such as age, genetics and gender (9, 34), as well as modifiable factors such as muscle structure, function and strength (35-37), fatigue and training history (9, 34) to name a few. Tissue capacity may also be influenced by the physiological response to psychological and sociocultural influences. For example, psychological factors such as stress and anxiety as well as personality traits such as obsessive passion, may lead to behavioural responses such as excessive training behaviours, disregarding recovery and ignoring minor injuries, which may in turn have a negative impact upon tissue capacity (38, 39). Psychological factors such as these, may also be influenced by sociocultural factors, such as exposure to stressful life events, coping mechanisms and presence of, or lack of, social support (18, 40). Therefore, clinicians should acknowledge the complex interaction between multiple different factors when assessing tissue capacity.

1.3.2 Cumulative tissue load

Historically, training errors have been cited as the main cause of running related injury amongst both runners (41) and researchers (1, 23). It is thought that greater training loads influence cumulative tissue loading by imposing a greater frequency of stress application to the musculoskeletal system. Once a cumulative tissue load has been reached that exceeds tissue capacity, injury occurs. Consequently, research has focused on identifying training parameters that may explain running injury development (23, 42).

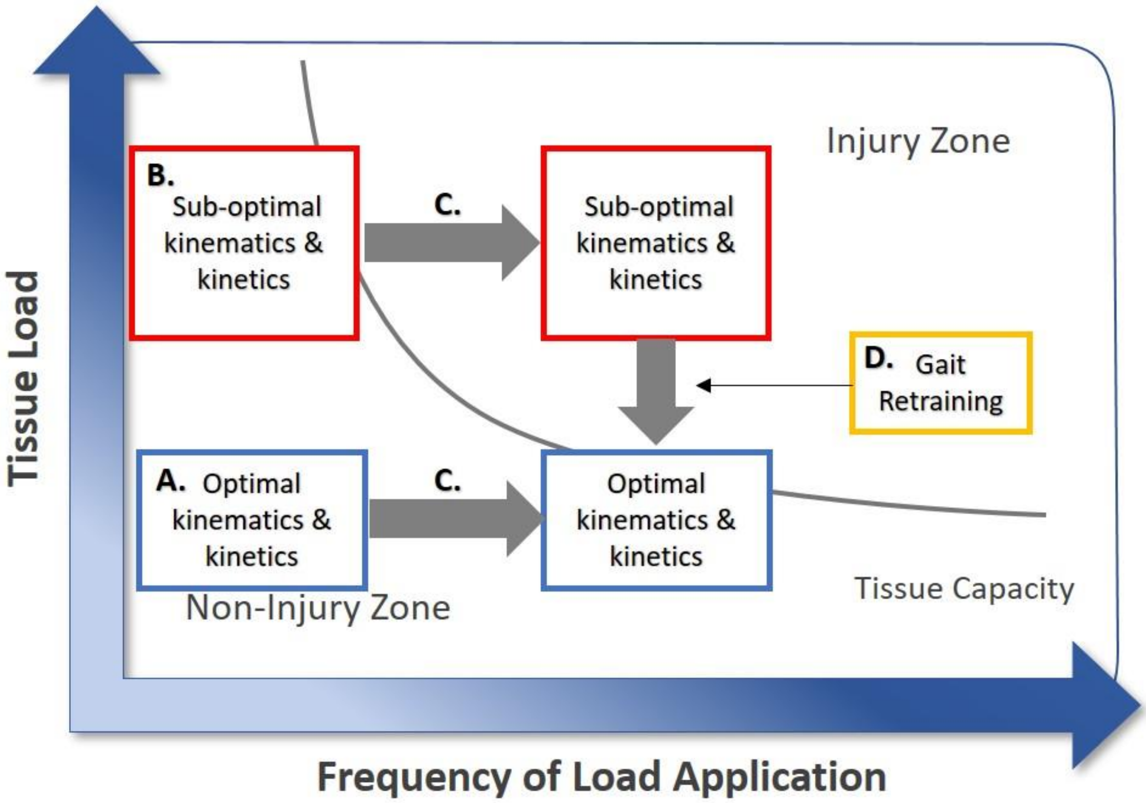
Despite the attempt to identify training errors associated with common running injuries, no conclusive evidence exists regarding the role of training errors within running related injuries. Some studies have reported higher weekly training volumes (43-45) or sudden increases in training volumes (46, 47) to be associated with an increased risk of sustaining a running related injury (43-45). However, several other studies have reported no differences in injury risk between runners completing high and low weekly running volumes (7), or following sudden increases to training volume (48). Interestingly a recent systematic review from Damsted et al (42) concluded that there is limited evidence to suggest sudden changes in training parameters are associated with an increased injury risk. This raises questions as to why some runners are able to attain high weekly training volumes without sustaining an injury, while others are not.

One explanation may be due to the interaction effects between training load exposure and factors influencing tissue load. According to Bertelsen's model of running injury aetiology (28), cumulative tissue load is considered the sum of the tissue specific loads experienced per stride and the frequency of load application (Figure 1B). While training load exposure may explain the frequency of load application, the load applied per stride depends upon a variety of biomechanical, anatomical and training factors influencing the magnitude of the biomechanical loads and the structures they are applied to. Such factors include, but are not limited to, body mass, joint congruence, running speed, training surface, running shoes, kinematics and kinetics; all of which influence the specific tissue structures loaded and the magnitude of loads encountered (28). Therefore, factors influencing tissue load per stride should be considered alongside training load exposure, as the interaction between the two will have a direct impact upon cumulative tissue loading.

1.3.3 Running biomechanics: influencing tissue loading

Running biomechanics are one factor influencing the load or stress placed upon musculoskeletal structures during each loading cycle of a run (1, 28, 49). Several biomechanical studies have reported kinetic and kinematic parameters to directly influence the load or stress encountered by muscles, bones and joints. For example, the kinematic parameters of hip internal rotation (50), hip adduction (50) and knee flexion (51) have all been shown to influence patellofemoral joint stress during running. Similarly, kinetic parameters such as vertical loading rate are thought to influence tibial bone loading (52, 53). By imposing greater loads upon the musculoskeletal structures per stride of a run, biomechanical parameters could cause an individual to function closer to their tissue capacity (Figure 2).

Figure 2: Adapted stress frequency curve from Hreljac & Ferber (1).
A = an individual with optimal kinematics & kinetics experiences a low tissue load per foot contact. B = an individual with sub-optimal kinematics & kinetics experiences a higher tissue load per foot contact, however, may not exceed tissue capacity if frequency of load application is low. C = as the frequency of load application increases, the cumulative tissue load may cause individual B to exceed their tissue capacity while individual A remains within their tissue capacity. D = Gait retraining aims to change sub-optimal kinematics & kinetics to optimal, reducing the tissue load per contact allowing an individual to function within their tissue capacity.



In the current literature several prospective and retrospective studies support the link between running mechanics and common running related injuries, including Achilles tendinopathy (54, 55), medial tibial stress syndrome (55, 56), patellofemoral pain (57, 58) and iliotibial band syndrome (26, 59). Interestingly many of these studies report similar kinematic and kinetic parameters across multiple different running related injuries, for example, hip adduction has been associated with MTSS (56), ITBS (26) and PFP (58), while increased vertical loading rate (53) and peak horizontal breaking force (60) has been associated with global injury development amongst runners. This suggests that there may be kinematic and kinetic parameters which are associated with global running injury; increasing tissue load per foot contact of a run. Identification of such parameters would be invaluable to clinicians, as it could allow the development of rehabilitation and prehabilitation interventions specifically targeted at the underlying mechanics.

Importantly, based on several injury causation models, biomechanics alone may not be enough to cause injury if there is limited exposure to external training load, as the cumulative tissue load may not be enough to exceed tissue capacity (28, 30, 33, 61). Running biomechanics are one factor that can increase the loads applied to the musculoskeletal system per foot contact of a run, whereas the frequency this load is applied can be influenced by the number of loading cycles imposed by external training load. The interaction between the two can subsequently influence the cumulative loads applied to the musculoskeletal system across a run or training week (Figure 1B) and thus whether this load exceeds the tissue capacity of an individual (Figure 1D).

Using an adapted stress frequency curve as an illustrative example (Figure 2), if a runner has sub-optimal biomechanics and runs a relatively low frequency, the tissue load may be elevated, but they may not exceed their tissue capacity as the cumulative load remains relatively low (Figure 2B). However, if this same individual increases their training volume, the frequency of load application may result in a cumulative tissue load that exceeds tissue capacity and injury occurs (Figure 2C). Conversely, a runner with optimal biomechanics may be able to safely increase their training load as the tissue load per stride, and subsequently the cumulative tissue load, remains low and does not exceed tissue capacity (Figure 2C). This interaction may explain why some runners can attain high weekly training volumes while others become injured. If such an interaction exists, it would be important to consider the training load a runner is exposed to and whether their mechanics may limit their ability to increase these training loads without injury development.

1.3.4 Targeting Running Mechanics in the Rehabilitation Process

If specific running mechanics are associated with common running related injuries, clinical interventions that target these mechanics may reduce tissue loads allowing runners to recover from injury and increase their training volumes. In the current literature, several studies have utilised strength interventions with the aim of improving running kinematics (62-65). However, despite idenitifying significant imporvements in strength, no differences in kinematics have been observed. In two separate studies, the single leg squat exercise was used with the aim of imporving frontal plane hip and pelvis angles (64, 65). These studies reported significant imporvements in both hip adduction and contralateral pelvic drop during the single leg squat, however this was not transferred to running. This suggests that strength training alone is insufficient to improve running kinematics and that and that task specific movement retraining may be a more effective intervention.

1.3.5 Gait Retraining: Lowering Tissue Stress in Rehabilitation

Gait retraining has been proposed as a movement specific intervention aimed at correcting sub-optimal running mechanics. Gait retraining is the process of altering running technique or specific movement patterns using internal and/or external cues (49, 66). Once the individual has learnt the desired running technique, the aim is to then to reinforce and maintain the learnt running mechanics. There are several different methods of gait retraining identified in the current literature, these include foot strike manipulation, visual feedback, step width modification and step rate modification (67-71). Several studies have shown running retraining can result in significant changes to joint specific movement patterns (69-72). For example, transitioning to a forefoot running pattern has been shown to reduce stance phase knee flexion range of movement (73-75).

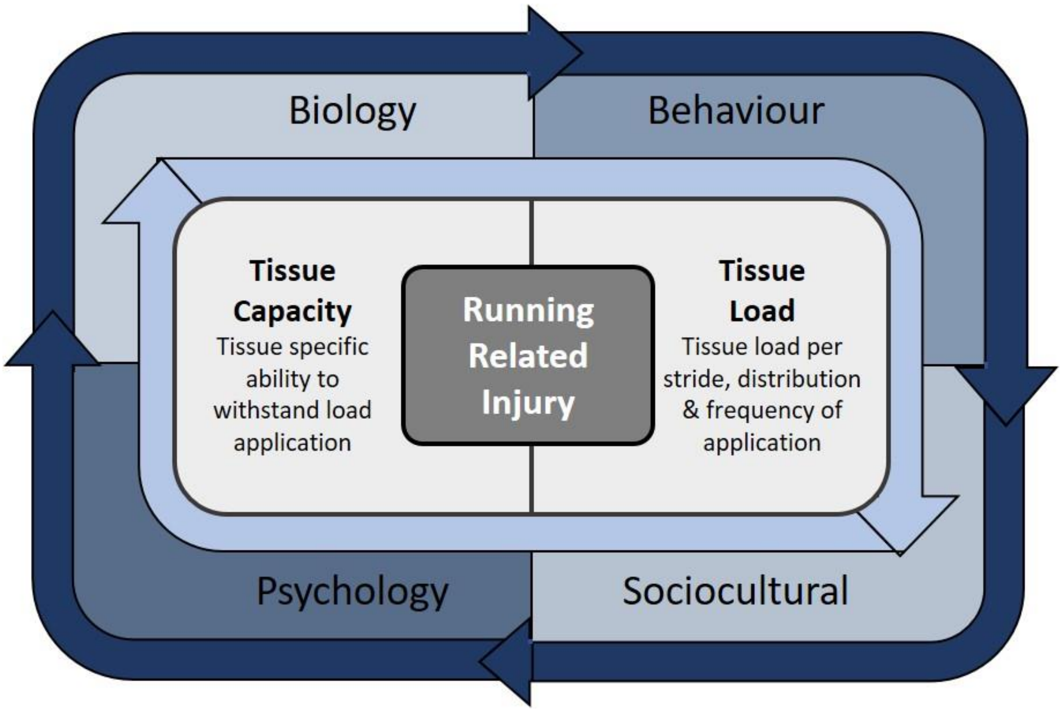
It is thought that modifying running mechanics through gait retraining serves to reduce or redistribute load applied to the musculoskeletal system during each foot contact and subsequently the cumulative loads across a given run (Figure 2D). Evidence to support this idea comes from several studies investigating the effects of gait retraining on patellofemoral joint loads. Using healthy runners, Lenhart et al (51) reported that a 10% increase in running step rate resulted in a 14% reduction in peak patellofemoral force per loading cycle. Similarly, Willson et al (76) reported a 10% increase in step rate resulted in a 22.2% decrease in patellofemoral joint stress per step and a 7.5% decrease across a mile run. This supports the idea that modifying running mechanics through targeted interventions may reduce both the tissue load per stride and the cumulative tissue load across an entire run or weekly training period. Therefore, gait retraining could be a clinically effective intervention that may reduce the tissue loads imposed by biomechanical parameters, reducing the cumulative tissue load encountered during running and facilitating return to run amongst injured runners by allowing them to function within their injury threshold (Figure 2D).

1.3.6 Psychosocial considerations: the wider context

It is important to note that running biomechanics form just one aspect of the multifactorial construct of tissue load and clinically should be acknowledged as such. Additional factors may also influence tissue load, such as anatomical structure, running surface and running shoes, which can all influence how forces are distributed to tissues, as well as running speed, surface gradient and body mass which may influence the magnitude of the applied loads. Similarly, tissue loads can extend beyond those that are biomechanical to psychological loads as well (18, 32).

The injury framework proposed by Bertelsen et al (28) allows for the conceptualisation of running injury aetiology. However, from a clinical perspective, the framework should be taken into consideration amongst a much wider biopsychosocial model reflecting the complexity of sport injury (18, 19, 27). In particular, not only is there a dynamic relationship between tissue capacity and tissue load, but these constructs may be mediated by the interaction between a variety of psychological, biological, sociocultural and behavioural factors (Figure 3) (18, 19, 28). For example, sociocultural factors such as negative life events may influence psychological stress and mood status, which may in turn impact upon biological processes influencing tissue recovery and function (40, 77). Similarly, poor psychological coping mechanisms or obsessive personality traits may influence behavioural responses leading to elevated training loads and disregard for recovery (38-40, 77). Consequently, the interaction between such factors may mediate tissue capacity and tissue loading either directly or indirectly via other mediators (Figure 3).

Figure 3: Conceptual model of the biopsychosocial influences for running related injury. Adapted from the biopsychosocial model of injury by Wiese-Bjornstal (18) and conceptual framework of running injury aetiology proposed by Bertelsen et al (28). The inner layers represent the balance between tissue capacity and tissue load and their influence on running related injury. The outer layers signify the interaction between biological, sociocultural, psychological and behaviour factors which may have a mediating effect upon the inner layers. The surrounding arrows between each layer represents the dynamic relationship between constructs and their mediating effect on one another.



Not only could the interaction between biopsychosocial factors influence injury development but may also influence perceived injury severity and rehabilitation outcomes (18, 78, 79). Forsdyke et al (79) recently highlighted the influence of psychosocial factors on rehabilitation outcomes. According to their systematic review, feelings of anxiety, low self-confidence, low mood, fear of reinjury, poor coping and loss of social support were all associated with poor rehabilitation outcomes (79). Indeed, psychosocial factors such as these are also known to influence perceptions of injury severity (80). This could subsequently lead to maladaptive behaviours such as pain avoidance and fear of movement, which could in turn compound biological deconditioning of tissues negatively impacting return to sport outcomes (18, 78-80). Therefore, for optimal clinical outcomes, clinicians should acknowledge the complexity of injury causation and rehabilitation outcomes beyond the immediate set of risk factors. This is because running injury and rehabilitation outcomes are likely the result of a complex interaction between multiple determinants (18, 19, 27).

1.3.7 Summary, Aims & Objectives

The aim of the introduction was to highlight the role of running biomechanics as a singular risk factor for running injury aetiology amongst the much wider context of injury causation. From a clinical perspective, it is important to acknowledge the complex multifactual nature of running related injuries while still enhancing our understanding of the singular components that influence the larger picture. Bertlesen's (28) running injury framework allows for the conceptualisation of running biomechanics as one factor influencing the load or stress encountered by musculoskeletal tissues during running, which combined with an exposure to external training load, may influence cumulative tissue load and injury. In such cases, gait retraining interventions which modify running biomechanics, could reduce tissue stress per foot contact and cumulative tissue loading, assisting in the rehabilitation of injured runners. For clinicians, understanding the biomechanical contributors to running related injuries, how they may interact with training load exposure and the effect of gait retraining interventions, may enhance clinical assessment and management strategies directed towards this specific injury risk factor.

Therefore, the overarching aim of this thesis was to first identify biomechanical characteristics associated with common running injuries and explore whether training load exposure influences running kinematics, discussing the potential implications for injury development. Finally, this thesis aims to investigate whether gait retraining can be used to effectively improve biomechanics, clinical and functional outcomes amongst injured runners.

In order to achieve this, a narrative literature review was first conducted in order to identify gaps within the current literature and form specific research aims, objectives and hypotheses. These are presented in Section 2.5.1. These aims and objectives are addressed within subsequent chapters of the thesis. For the narrative literature review, the following aims and objectives were established:

1. Explore the literature to identify the kinematic and kinetic characteristics of common running related injuries, the reliability of kinematic assessment measures,

whether training load exposure influences injury risk and running kinematics and whether gait retraining interventions can effectively target running kinematics (Chapter 2). The specific objectives to achieve this aim were to:

a. Review and critically appraise the current literature investigating kinematic and kinetic characteristics of common running injuries in order to identify kinematic and kinematic parameters associated with common running injuries.
b. Review the current literature in order to establish the reliability and repeatability of kinematic measurements during running.
c. Review the current literature to identify whether training errors are associated with running injury development and whether running kinematics are influenced by the training loads runners are exposed to.
d. To review and critically appraise the literature reporting the effects of gait retraining interventions upon running kinematics and clinical outcomes amongst injured runners.

Impact: the overall impact of this narrative literature review was to provide a broad overview of what is currently known regarding the association between running biomechanics and running related injuries, how kinematics and running related injuries are influenced by training load exposure and the effect of gait retraining interventions upon specific running kinematics. Through achieving this aim, gaps within the current literature were identified in order to inform specific research objectives outlined in Section 2.5.1 and addressed within subsequent chapters of the thesis.

2 Chapter 2: Literature Review

2.1 Biomechanical characteristics of running related injuries

Despite the growing popularity of recreational running there is considerable risk of musculoskeletal injury, with approximately 50% of runners injured annually (11). Of all running related injuries, four of the most frequently cited include patellofemoral pain (PFP), iliotibial band syndrome (ITBS), medial tibial stress syndrome (MTSS) and Achilles tendinopathy (AT) (10, 12, 13), with incidence and prevalence rates reported to be as high as 20.8% and 22.7% for PFP (14), 9.1% and 12% for ITBS (12, 15), 20% and 9.5% for MTSS (12, 16) and 10.9% and 18.5% for AT (12).

Although running related injuries have a complex multifactorial aetiology, running biomechanics are cited as one injury risk factor. As presented within the introduction Chapter, running biomechanics are thought to influence the tissue load encountered per foot contact of a run (28, 61). When combined with an exposure to external training load this may influence the cumulative tissue load encountered during a single run and across a training week (28). If this cumulative tissue load exceeds tissue capacity, tissue damage may occur leading to the development of running related injuries.

Within current literature several studies have identified an association between running kinematics and Achilles tendinopathy (54, 55), medial tibial stress syndrome (55, 56), patellofemoral pain (57, 58) and iliotibial band syndrome (26, 59). Identifying the biomechanical characteristics associated with common running related injuries may provide clinicians with an understanding of the biomechanical contributors to running related injuries, for which subsequent rehabilitation interventions can be targeted towards.

Therefore, the objective of this first part of the literature review is to review and critically appraise the current literature investigating kinematic and kinetic characteristics of common running injuries in order to identify kinematic and kinematic parameters associated with common running injuries.

In the sections below, the literature comparing kinematic and kinetic characteristics of four common running injuries, Achilles tendinopathy, medial tibial stress syndrome, patellofemoral pain and iliotibial band syndrome are discussed separately. A visual overview of the literature and reported findings are provided in tables within each injury section in order to aid interpretation of the current literature investigating biomechanical characteristics of common running injuries and identify potential gaps in the current evidence. The literature review concludes by highlighting the common biomechanical patterns associated with running related injuries, identifying gaps and limitations to the current literature and discussing the idea that there may be common kinematic patterns associated with multiple different running related injuries.

2.1.1 Literature Search

In order to review the current literature electronic databases were searched in order to identify studies investigating kinematic and kinetic characteristics associated with common running related injuries. CINHAL, MEDLINE, SportDiscus and Web of Science were searched for all years up until March 2019. For each of the four running injuries investigated, pathology specific search terms were used and are presented in (Table 1). Following identification of relevant titles, abstracts were screened for relevance and full texts were then assessed against the below inclusion and exclusion criteria. References and citations of all included studies were searched to identify any additional studies which meet the inclusion/ exclusion criteria.

2.1.1.1 Inclusion

- Retrospective or prospective study design
- Include an injured population with a diagnosis of either AT, MTSS, ITBS or PFP
- Include a healthy control comparison of male or female runners
- Report kinetic, kinematic or spatiotemporal parameters

2.1.1.2 Exclusion

- Studies using military populations
- Studies that do not assess running
- Studies using fatigue protocols or test during prolonged running

- Conference abstracts

Table 1: Literature search: key terms and boolean operators.

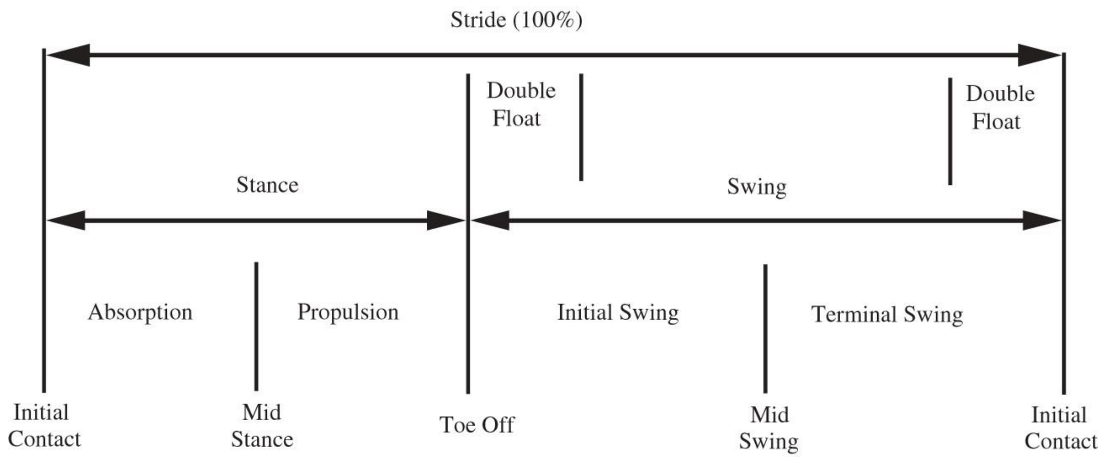
Search Terms
Iliotibial band syndrome OR Iliotibial band OR Iliotibial band friction syndrome OR ITBS
MTSS OR Medial Tibial Stress Syndrome OR Exercise related lower leg pain OR ERLLP OR Lower leg pain OR Tibial stress fracture OR Shin Splints
Patellofemoral pain OR Patellofemoral OR PFP OR PFPS OR Patellofemoral pain syndrome OR Chondromalacia patella
Achilles tendon OR Achilles OR Achilles tendinitis OR Achilles pain
The above pathology specific terms were combined with the below terms using the Boolean operator "AND"
Biomechanics OR kinetics OR kinematics
Running OR run OR jog OR runners

2.1.2 Key terms and definitions

2.1.2.1 The Gait Cycle

The gait cycle is the time period of events that occur during locomotion from when the foot first contacts the ground, to when that same foot contacts the ground again (81) (Figure 4). During running, the phases of the gait cycle can be broken down into the stance phase and the swing phase. The stance phase begins with initial contact, the point where the foot first contacts the ground, which is followed by a force absorption period. During the absorption period the body must decelerate the velocity of the centre of mass in both the vertical and horizontal direction(s) as it comes into contact with the ground (81), this is often referred to as the breaking phase peaking at mid stance. At mid stance the centre of mass reaches its lowest vertical point in the gait cycle and is positioned directly over the centre of pressure. Following mid stance is the propulsion phase, where the body must accelerate the centre of mass upwards and forwards, terminating at toe off as swing phase begins.

Figure 4: The running gait cycle cited from Dugan (82). Figure highlights the different phases of the gait cycle dividing the gait cycle into the stance phase and swing phase.



During initial swing there is a period where both feet are no longer in contact with the ground, referred to as double float (Figure 4), where the trailing leg begins to swing forwards and the opposite leg reaches terminal swing in preparation for initial ground contact (81, 82). Once the trailing leg passes the midline of the body it then begins to reach terminal swing where the limbs begin to prepare for initial contact and the beginning of a second stance phase and second gait cycle.

2.1.2.2 Kinetics

Kinetics is the measurement of forces acting on the body during running (83). This can include internal forces, such as those created by muscles, tendons and ligaments, and external forces both acting on the body as a whole, or those acting on individual joints (84).

The ground reaction force is the external force exerted by the ground acting on the body as a whole, as it contacts the ground. It can be separated into three main components, the horizontal component, vertical component and mediolateral component, each representing the direction in which the force is applied.

Joint moments represent the rotational force acting across a joint and are calculated using both external forces and joint kinematics (81). As external joint moments represent the forces acting on a joint, this information can provide an indication of internal forces required for to resist or counteract these forces, the internal joint moment (85). In relation to running injuries, greater joint moments could therefore represent greater loading of a joint or the surrounding muscles (85).

2.1.2.3 Kinematics

Kinematics is the description of joint movements and does not reflect the forces acting on the body (81). The focus of kinematics concerns the movement of specific joints of interest across three planes of movement, the sagittal, frontal and transverse planes. These movement patterns are most frequently measured as degrees of movement, either across the entire gait cycle, specific phases of the gait cycle such as mid stance, or at discrete points, such as peak joint angles during stance phase.

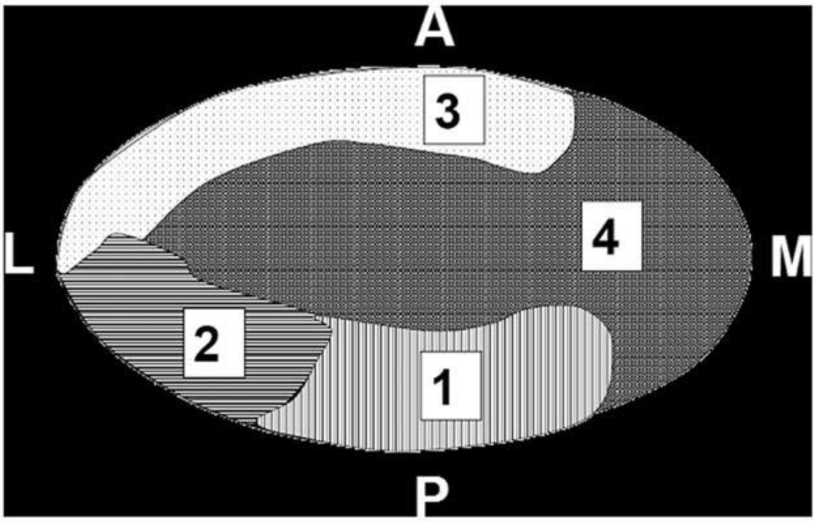
The stance phase of running is the period of the gait cycle where external and internal forces are at their greatest and joint angles can reach their maximum. As such, the musculoskeletal system is considered to be placed under considerable external and internal loads. This time point of recreational endurance running is possibly when the body is most vulnerable to injury. This could explain why the stance phase has seen the most research attention with respect to running injuries. This thesis will discuss the biomechanical parameters associated with common running related injuries. As the current literature focuses on the stance phase of running, it is this period of the gait cycle that focus will be driven towards.

2.1.3 Achilles Tendinopathy

2.1.3.1 Achilles Tendon Anatomy

The Achilles tendon is comprised of fascicles originating from the medial and lateral head of the gastrocnemius and the soleus muscle (86). The medial head of the gastrocnemius arises from the popliteal surface of the femur and the lateral head arises from the posterolateral aspect of the femoral condyle (87). The soleus arises from the medial border of the tibia and the posterior surface of the fibula (87). As the fibres of the Achilles tendon descend distally, they begin to spiral forming a helical twist before attaching to the calcaneus. The fibres of the medial gastrocnemius rotate laterally and posteriorly while the fibres of the lateral gastrocnemius rotate anteriorly (Figure 5) (88). The fibres of both the medial and lateral gastrocnemius appear to surround the soleus fibres, with the gastrocnemius fibres forming the superficial portion and Soleus forming the mid and deep portion of the Achilles tendon (Figure 5) (88). This complex anatomical arrangement, combined with independent function of the muscles comprising the Achilles tendon, is likely to influence the internal loading of the tendon; allowing for inter-fascicle sliding and non-homologous loading of the Achilles tendon which may have implications for pathology development.

Figure 5: Transverse cross-section through the left Achilles tendon,1cm above tuber calcanei reprinted from Szaro et al (88). (1) The fibers from the medial part of the medial head of the gastrocnemius, (2) the fibers from the lateral part of the medial head of the gastrocnemius,(89) the fibers from the lateral head of the gastrocnemius and(4) the fibers from the soleus, A: anterior, P: posterior, M: medial, L: lateral.



2.1.3.2 Pathomechanics of Achilles Tendiniopathy

Although the exact pathomechanics of Achilles tendinopathy are poorly understood (90-93), several biomechanical mechanisms have been suggested to contribute to pathology development. In 1984, Clement et al (94) proposed the “whiplash” theory, whereby excessive foot pronation and knee extension are thought to cause repeated “whipping” and “wringing” of the Achilles tendon, resulting in vascular impairment, micro-tearing and subsequent tendon degeneration. However more recently, theories have suggested that non-uniform displacements occurring between tendon fascicles (95-97), or elevated compressive and frictional forces between the adjacent plantaris tendon, may also play a role in tendinopathy development.

Several recent studies have reported the presence of non-uniform displacements occurring between tendon fascicles, suggesting this may form a mechanism triggering AT development (95-97). As the tendon is comprised of three distinct sub-tendons, each with a separate muscular origin, independent function of tendon portions may lead to a rise in interfascicle sliding and shear forces; causing damage to intermolecular cross links and leading to subsequent pathology development (92, 93, 95, 98). This non-uniform behaviour of tendon fascicles has been shown to be influenced by knee (99), ankle and foot kinematics (95, 96, 100-102) as well as muscle activation patterns within the triceps surae (97, 103). For example, in several separate studies, greater displacement of the mid and deep tendon relative to the superficial portion, has been observed as ankle dorsiflexion increases (95, 96, 100-102). Similarly, greater tendon displacements and strain has been observed within the soleus sub-tendon as knee flexion increases (99), as well as a strain increase of up to 15% observed within the medial aspect of the tendon with calcaneal eversion (104).

Differences in muscle activation patterns between the soleus, medial and lateral gastrocnemius are also thought to influence non-uniform loading within the Achilles tendon (97, 103, 105). In an anatomical study by Finni et al (97), stimulation of the different muscles comprising the triceps surae were observed to cause an increase in interfascicle displacements and strains within the Achilles tendon. This is further supported by a biomechanical modelling study by Handsfield et al (103), reporting up to 85% of the non-uniformity in tendon displacements to be explained by muscular forces exerted by the triceps surae. Considering muscle activation patterns of the triceps surae can vary with transverse plane foot rotation (106, 107), knee flexion (108) and rearfoot eversion (109), it is possible that running kinematics may contribute to altered muscular forces and tendon displacements.

Recent studies have also suggested the plantaris tendon may play a role in mid portion Achilles tendinopathy (110-113). Although the exact anatomical position of the plantaris can demonstrate considerable inter-individual variation (113, 114), the tendon is reported to arise from the lateral aspect of the supracondylar line of the femur, passing distally between the medial gastrocnemius and soleus muscles to insert on the medial aspect of the calcaneus (114). In some instances, the plantaris has been found to merge with the medial aspect of the Achilles tendon (113, 114). Biomechanical studies have reported the plantaris to be significantly stiffer and stronger than that of the Achilles tendon (111) and due to its anatomical position, has been shown to result in friction induced shear stress during repeated ankle plantarflexion (115), as well as elevated compressive forces with rearfoot eversion (116).

Based on the current literature it appears that there may be several biomechanical mechanisms that could drive Achilles tendinopathy. These mechanisms could lead to excessive torsional, shearing and/or compressive stress' placed upon the Achilles tendon. Subsequently contributing to repeat microdamage, triggering a pathological response and the onset of Achilles tendinopathy. Based on the proposed pathomechanical methods, runners with Achilles tendinopathy may be expected to demonstrate kinematic patterns at the knee and ankle such as increased knee and ankle dorsiflexion excursion, increased peak ankle dorsiflexion, increased peak knee flexion and increased rearfoot eversion.

2.1.3.3 Biomechanics in Achilles Tendinopathy

A visual summary of the number of studies and reported findings, investigating the difference in running kinematics and kinetics between runners with AT and injury free controls are presented in Table 2, Table 3, Table 4 & Table 5.

2.1.3.3.1 Kinematics

From a pathomechanical perspective, it is possible that increased ankle dorsiflexion and knee flexion during the stance phase of running could increase intra-tendinous shear stress and contribute to Achilles tendinopathy. However, based on current literature there is limited evidence to support such a link (Table 2). One study reported runners with AT to demonstrate increased peak ankle dorsiflexion and knee flexion angles during running (117), whereas further studies have failed to identify any significant difference in peak ankle dorsiflexion (118-121), dorsiflexion range of movement (118, 121) and peak knee flexion between runners with AT and controls (119, 120). One case control study reported reduced knee flexion range of movement amongst runners with AT (119) and one prospective study observed runners with AT to demonstrate reduced peak ankle dorsiflexion and knee flexion (54). However, the prospective study by Hein et al (54) contained a small sample size of only 9 subjects. Therefore, based on the current findings there appears to be limited evidence to support the link between sagittal plane knee and ankle kinematics and Achilles tendinopathy.

Table 2: Visual summary of the number of studies and reported findings, investigating the difference in distal running kinematics between runners with AT and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. RoM = Range of Movement.

			Significantly Increased	Significantly Reduced	Non-significant
Knee	Knee Flexion	Initial Contact			2 1
		Peak	1	1	2 1
		RoM		1	1 1
	Knee Adduction	Peak			
	Knee Abduction	Peak			
	Knee Internal Rotation	Peak		1	
	Knee External Rotation	Peak			
Ankle	Ankle Dorsiflexion	Initial Contact			2 1
		Peak	1	1	4 2
		RoM	1		2 2
Foot	Rearfoot Eversion	Initial Contact	2 1 1		
		Peak	1		5 2
		Time to Peak			3 1
		RoM	2		3 2
		Duration	1		
		Velocity			2 1
		Toe Off	1		

Similarly, there is currently a lack of evidence to support an association between rearfoot kinematics and AT (Table 2). Only one study in the literature has reported greater peak rearfoot eversion during running in subjects with AT (117), however this study selectively recruited subjects who demonstrated increased rearfoot eversion at baseline, thus confounding their results. Three studies have identified greater rearfoot eversion at initial contact (54, 120), one reported greater duration of eversion during the stance phase (55) and greater rearfoot eversion at toe off (55) and two studies reported greater eversion range of movement (120, 121) (Table 2). However, these studies should be interpreted in light of their limitations. Firstly, Ryan et al (121) had subjects run barefoot, which may not be a true representation of their normal runing gait, Hein et al, (54) were limited to a small sample size of only 9 subjects, and as mentioned earlier, Donoghue et al (120) selectively recruited subjects with increased rearfoot eversion at baseline. Combined with results from several further studies reporting no difference in peak eversion angles (55, 118, 120-122), maximal eversion velocity (55, 121) and eversion range of movement (55, 118, 121, 122), the evdience to support the pathomechanical model of intra-tendionous shear stress induced by rearfoot eversion, or Clement's theory of repeated "whipping" and "wringing", is currently lacking.

The lack of evidence to support the role of lower limb kinematics is also supported by the results of three systematic reviews with two meta analyses (123-125) (Table 2). Based on the pooled findings across multiple cross-sectional studies, Sancho et al (123) and Mousavi et al (124), concluded that there is limited to strong evidence for no between-group difference in peak ankle dorsiflexion, ankle dorsiflexion range of movement, peak rearfoot eversion and eversion range of movement. However, Sancho et al (123) did acknowledge that current evidence regarding rearfoot kinematics is conflicting, due to the results of one study from Becker et al (55) suggesting there is some, albeit limited, evidence that rearfoot eversion at toe off and duration of rearfoot eversion is increased in runners with AT.

Some very limited evidence suggests there could be a link between proximal kinematics and Achilles tendinopathy. In a study by Williams et al (126) they reported a group of runners with AT to demonstrate reduced knee internal rotation during stance. On inspection of their data, this appeared to be the result of a greater femoral internal rotation on the tibia resulting in the appearance of reduced knee rotation. Results from Creaby et al (118) also point to potential proximal influences in AT, reporting hip internal rotation at peak vertical ground reaction force, to be 7.34^{o} greater amongst runners with AT compared to controls. Although this failed to reach statistical significance, there was a moderate effect size suggesting there may be potential proximal influences. Combined with some limited evidence reporting delayed onset and shorter activation periods of both gluteus medius and maximus amongst runners with AT (123), proximal influences for AT may warrant further research. Interestingly, there is a significant lack of evidence investigating hip, pelvis and trunk kinematics in AT with no current study reporting trunk or pelvis kinematics (Table 3).

Table 3: Visual summary of the number of studies and reported findings, investigating the difference in proximal running kinematics between runners with AT and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. RoM = Range of Movement.

			Significantly Increased	Significantly Reduced	Non-significant
Trunk	Forward Lean	Initial Contact			
		Peak			
	Ipsilateral Trunk Lean	Initial Contact			
		Peak			
Pelvis	Anterior Pelvic Tilt	Initial Contact			
		Peak			
	Contralateral Pelvic Drop	Initial Contact			
		Peak			
Hip	Hip Flexion	Initial Contact			1
		Peak			1
	Hip Adduction	Initial Contact			
		Peak			1
		RoM			1
	Hip Internal Rotation	Initial Contact			
		Peak (at VGRF peak)			1
		RoM			

2.1.3.3.2 *Kinetics*

Limited studies have investigated kinetic associations to Achilles tendinopathy (Table 4 &Table 5). Of the available evidence, several studies have reported no difference between runners with AT and controls for peak vertical ground reaction force (55, 122, 127), vertical impact peak (119, 122, 127), time to vertical impact peak (122) and vertical loading rate (119, 122) as well as both horizontal and frontal plane ground reaction force parameters (55, 119, 122, 127). These findings are further supported by the results from two systematic reviews, reporting limited evidence for no significant difference in vertical and horizontal ground reaction force profiles between runners with AT and controls (123, 128).

With regards to joint moments, only two separate studies have reported lower limb joint moments amongst runners with AT (118, 126) (Table 5). One study reported increased peak hip external rotator moment and hip adduction moment impulse (118) and another study reported reduced peak tibial external rotator moment (126), however the latter study was conducted in a small sample of only 8 runners who were asymptomatic at the time of testing. Therefore, based on these findings as well as the findings of a recent systematic review (123), there is currently limited evidence linking kinetic parameters to AT.

Table 4: Visual summary of the number of studies and reported findings, investigating the difference in ground reaction force profiles during running between runners with AT and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. GRF = Ground Reaction Force.

		Significantly Increased	Significantly Reduced	Non-significant
Vertical GRF	Peak Vertical Ground Reaction Force			3 2
	Vertical Impact Peak			3 1
	Time to Vertical Impact Peak (s)			1
	Vertical Loading Rate (BW/s)			2 1
Horizontal GRF	Peak Breaking			3 1
	Time to Peak Breaking			1
	Peak Propulsive			4 1
	Time to peak Propulsive			1
Frontal GRF	Peak Medial			1
	Peak Lateral			

Table 5: Visual summary of the number of studies and reported findings, investigating the difference in lower limb kinetics during running between runners with AT and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings.

		Significantly Increased	Significantly Reduced	Non-significant
Hip Kinetics	Peak Flexor Moment			1
	Peak Adduction Moment			1
	Adduction Moment Impulse	1		
	Peak External Rotation Moment	1		
Knee Kinetics	Peak Extensor Moment			
	Peak Abductor Moment			
	Abduction Moment Impulse			
	Peak External Rotation Moment			1
	Peak Tibial External Rotation Moment		1	
Ankle Kinetics	Peak Plantarflexor Moment			1
	Peak Inversion Moment			1

2.1.3.4 Summary and limitations to the current literature

Currently there is limited evidence supporting a link between altered running kinematics and AT. Although multiple biomechanical theories have been proposed to contribute to the development of AT, there is currently limited kinematic or kinetic evidence to support such a link. However, there are several limitations to the current literature which should be acknowledged. Firstly, the modelling techniques used to track rearfoot kinematics may not provide an accurate representation of true foot movement. Most studies calculate rearfoot kinematics using markers attached directly to the heel of the shoe, which may not be an accurate representation of the underlying movement of the foot (129-131). Sinclair et al (131) investigated the difference between foot mounted and shoe mounted markers on foot kinematics identifying shoe mounted markers to significantly underestimate frontal plane foot kinematics. Therefore, it is possible that the lack of a clear link between rearfoot eversion and AT is due to the poor accuracy of kinematic measurements of the rearfoot. Second, a current gap within the literature is the distinct lack of studies investigating associations between proximal kinematics and AT (Table 3). Recent studies have shown the existence of a kinematic coupling between proximal and distal segments which may have an impact upon lower limb tissue stress. Specifically, studies have identified correlations between hip adduction, hip internal rotation and rearfoot eversion during running and walking (132-134). This suggests that aberrant proximal kinematics may influence foot and lower limb function. Frontal plane kinematics of the pelvis may also have an impact upon lower limb mechanics and tissue stress. As the pelvis drops away from the stance leg, there is medial shift in the centre of mass, which may contribute to altered foot pressure distribution and/or compensatory mechanics at the lower limb (135, 136). However, to date, no study has reported frontal plane pelvis kinematics amongst runners with AT.

Currently, only one study has reported hip kinematics in runners with AT, suggesting a possible link between proximal kinematics and AT may exist (118). This is further supported by data from electromyographic studies identifying runners with AT to demonstrate delayed onset of gluteus maximus and gluteus medius muscle activity (137). Neuromuscular deficits at the hip have previously been linked to increased hip adduction excursion angles (138), suggesting that there may be associations between aberrant hip and pelvis kinematics and runners with AT that has not yet been investigated. Therefore, future should consider the role of proximal kinematics, particularly hip and pelvis kinematics, amongst runners with AT.

2.1.4 Medial Tibial Stress Syndrome

2.1.4.1 Pathomechanics of Medial Tibial Stress Syndrome

Medial tibial stress syndrome (MTSS) presents as an exercise induced pain syndrome along the posteromedial border of the tibia (139). The exact pathology is still debated with theories suggesting MTSS could be a fascial traction injury (140, 141) or a bone overload injury (142). It is possible that MTSS could represent a biomechanical overload syndrome, ranging along a clinical spectrum from fascial traction and tendinopathy, periosteal oedema, periosteal remodelling, tibial bone stress reaction and tibial stress fracture (140, 141, 143-145). Therefore, these stages of pathology may all be considered as part of a MTSS continuum (144). It is thought that biomechanics increasing the load placed on the medial border of the tibia can contribute to gradual overload of the medial aspect of the tibia, resulting in pathology development.

One mechanism of tibial overload is said to occur through increased traction of the crural fascia upon the medial tibial periosteum (140, 141, 145). It is possible that ankle invertor muscle activity may increase tension through the crural fascia, placing the medial border of the tibia under excessive traction stress, resulting in repeated microdamage to the periosteum (140, 141, 145). To investigate this idea Bouche & Johnson (145) increased the tension of the ankle invertors of cadavers while measuring periosteal strain at the medial tibia. As they increased tension of the tibialis posterior, flexor digitorum longus and soleus they found periosteal strain to increase linearly which they reported to be due to increased tension through the crural fascia. Given this finding, we may expect to observe kinematic patterns in people with MTSS which influence increased muscular work of the invertors. For example, it is possible that increased rearfoot eversion or internal rearfoot inversion moment, could create the need for an increase in biomechanical demand of the ankle invertors leading to overload of the medial tibia.

Another mechanism of MTSS development is that of abnormal bending, torsion and shear forces placed on the tibia. Long bones such as the tibia can withstand large amounts of compressive forces, however their ability to withstand torsional and shear stress is considerably lower (146). Subsequently abnormal forces are thought to cause excessive microdamage to the tibia and ultimately lead to tissue failure (147). George & Vashishth (146) investigated the effects of axial and torsional loading on the fatigue life of bovine tibias. They reported that combined torsional and axial loading resulted in a seven-fold decrease in the fatigue life of the tibia. Furthermore, in a histological study of tibia biopsies, Winters et al (148) identified the presence of microcracks, suggestive that abnormal biomechanical loading of the tibia results in overload and injury. Torsional loading to the tibia is likely to be influenced by frontal and transverse plane kinematics at the foot, as well as proximal segments such as the hip, pelvis and trunk. It would therefore be hypothesised that runners with MTSS, may demonstrate abnormal forces in the medial to lateral direction and abnormal kinematics in the frontal and transverse planes at both distal and proximal segments. The kinematic and kinetic evidence supporting these ideas are discussed below.

2.1.4.2 Biomechanics in medial tibial stress syndrome

As MTSS can be considered a biomechanical overload syndrome to the medial boarder of the distal third of the tibia and the lack of clear diagnostic criteria for MTSS (139), this literature review included biomechanical studies of runners with a retrospective history of tibial stress fracture. This was deemed necessary due to the current lack of studies reporting the biomechanical characteristics of runners with a current diagnosis of MTSS (n = 4) (55, 149-151). Table 6, Table 7 and Table 8 provide a visual summary of the number of studies and reported findings, investigating the difference in running kinematics and kinetics between runners with MTSS and injury free controls.

2.1.4.2.1 Kinematics

Increased rearfoot eversion may contribute to both muscular traction at the medial tibia and abnormal torsional loading. This idea is supported by research from several case-control kinematic studies, including two prospective studies (149, 151) (Table 6), identifying runners with a history of MTSS to demonstrate increased peak rearfoot eversion (56, 149, 152), a more everted foot at toe off (55) and greater duration of rearfoot eversion during the stance phase (55, 151). As the ankle invertors are required to control eversion of the rearfoot, increased rearfoot eversion may lead to a greater biomechanical demand on the invertor muscles and a corresponding increase in muscular traction at the medial tibia. Repeated over several loading cycles, this may lead to overload of the periosteal boarder of the tibia and the development of MTSS.

Table 6: Visual summary of the number of studies and reported findings, investigating the difference in distal running kinematics between runners with MTSS and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. RoM = Range of Movement.

			Significantly Increased	Significantly Reduced	Non-significant
Knee	Knee Flexion	Initial Contact			2 (green)
		Peak		1 (green)	1 (green), 1 (red)
		RoM			2 (green), 1 (red)
	Knee Adduction	Peak			2 (green), 1 (red)
	Knee Abduction	Peak			
	Knee Internal Rotation	Peak	1 (green)		2 (green), 1 (red)
	Knee External Rotation	Peak			
Ankle	Ankle Dorsiflexion	Initial Contact			1 (red)
		Peak			1 (red)
		RoM			1 (red)
Foot	Rearfoot Eversion	Initial Contact			2 (red)
		Peak	2 (green), 1 (red)		1 (green), 1 (red)
		Time to Peak	1 (red)		
		RoM			1 (green), 1 (red)
		Duration	1 (green), 1 (red)		
		Velocity			1 (green), 1 (red)
		Toe Off	1 (red)		

Rearfoot eversion may also influence torsional loading of the tibia through dynamic coupling with the tibia. As the calcaneus everts during stance, the close articulations between the subtalar joint and tibia result in a concurrent increase in subtalar joint pronation and tibial internal rotation (153). As such, kinematic studies have shown positive correlations between rearfoot eversion and tibial internal rotation during running (132, 153). As knee kinematics are the result of tibia motion relative to the femur, this may result in an increase in transverse plane knee motion, leading to elevated torsional forces at the tibia. However, in the current literature only one study has reported runners with a history of MTSS to demonstrate this coupling effect between increased rearfoot eversion and internal rotation of the tibia during running, resulting in increased peak knee internal rotation (56). A further three studies have failed to identify any difference in transverse plane knee kinematics between MTSS runners and controls (151, 152, 154). Therefore, current evidence does not appear to support an association between knee kinematics and MTSS (Table 6).

Kinematics at proximal segments such as the hip, pelvis and trunk may also contribute to abnormal loading placed on the tibia and influence distal kinematics at the foot. Two studies have reported increased hip adduction angles in runners with MTSS (56, 152) with one additional study reporting increased hip internal rotation (150) (Table 7). It is thought that hip adduction and internal rotation will alter the load distribution through the lower limbs, increasing the torsional forces placed on the tibia (56, 152). Due to the tight articulations between the rearfoot and tibia, as well as the tibia and femur, it is also possible that hip kinematics could have a direct impact upon rearfoot kinematics (155). Luz et al (132) investigated this link between proximal and distal kinematics reporting positive correlations between hip adduction and rearfoot eversion during running (132). However, there is conflicting evidence from one prospective study and two case-control studies, who failed to identify any significant difference in peak hip adduction (149, 154) and peak hip internal rotation (149, 152, 154) (Table 7). Therefore, although there is a theoretical link between hip kinematics and lower limb tissue stress, there is only a limited number of studies reporting hip kinematics amongst runners with MTSS, with no systematic review formally investigating the role of kinematics in MTSS. Consequently, further evidence is needed to support the association between hip kinematics and MTSS.

Table 7: Visual summary of the number of studies and reported findings, investigating the difference in proximal running kinematics between runners with MTSS and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. RoM = Range of Movement.

			Significantly Increased	Significantly Reduced	Non-significant
Trunk	Forward Lean	Initial Contact			
		Peak			
	Ipsilateral Trunk Lean	Initial Contact			
		Peak			
Pelvis	Anterior Pelvic Tilt	Initial Contact			
		Peak			
	Contralateral Pelvic Drop	Initial Contact			1 (red)
		Peak	1 (green) 1 (red)		
Hip	Hip Flexion	Initial Contact			
		Peak	1 (green)		
	Hip Adduction	Initial Contact			1 (red)
		Peak	2 (green)		1 (green) 1 (red)
		RoM			
	Hip Internal Rotation	Initial Contact			1 (red)
		Peak	1 (green)		2 (green) 1 (red)
		RoM			

Movements of the pelvis are also likely to have an influence on lower limb mechanics, influencing MTSS development. Two studies have identified increased contralateral pelvic drop amongst runners with MTSS (149, 150) (Table 7). It is possible that contralateral pelvic drop my influence MTSS through compensatory lower limb kinematics or by altering the force distribution through the lower limbs. Lin et al (156) reported that contralateral pelvic drop is the kinematic parameter which most strongly influences medio-lateral displacement of the centre of mass. As the pelvis drops away from the stance limb, there is a medial shift in the centre of mass (136). Consequently, lower limb pressure distribution is likely to be altered and compensatory kinematics, such as increased hip adduction or ipsilateral trunk lean may occur in order to maintain balance (135, 136, 157). Therefore, elevated contralateral pelvic drop could conceivably impact MTSS by increasing torsional and bending forces on the tibia, as the centre of mass shifts in the medio-lateral direction and/ or through the influence upon subsequent hip and foot kinematics.

2.1.4.2.2 Kinetics

Several studies have focused on ground reaction force parameters, suggesting a link between vertical loading rates and tibial stress fracture (Table 8). Vertical loading profiles are thought to provide a measure of the magnitude or "dose" of loading applied to the tibia, with elevated loading rates representing the speed in which impact loading forces are applied. It is thought that due to the viscoelastic properties of musculoskeletal structures, elevated loading rates may lead to earlier tissue fatigue and failure (52, 53). Several studies provide evidence to support a link between elevated vertical loading rates in runners with a history of tibial stress fracture (Table 8), including three retrospective case control studies (52, 56, 158) and the pooled results from two meta-analysis (128, 159).

Additionally, a number of studies have reported MTSS subjects to demonstrated elevated peak positive tibial accelerations (52, 56, 160) and elevated free moment (56, 161). Free moment represents the rotational force required to resist adduction or abduction of the foot relative to the ground (162), with elevated free moment suggested to represent greater torsional loading applied to the lower limb during stance. The magnitude of free moment is directly influenced by the magnitude of foot pronation (162), further supporting the link between rearfoot eversion and torsional loading of the lower limb.

Table 8: Visual summary of the number of studies and reported findings, investigating the difference in ground reaction force profiles and lower limb kinetics during running between runners with MTSS and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. GRF = Ground Reaction Force.

		Significantly Increased	Significantly Reduced	Non-significant
Vertical GRF	Peak Vertical Ground Reaction Force			3 2
	Vertical Impact Peak			3 1 2
	Time to Vertical Impact Peak			2 1
	Vertical Loading Rate	3 2		1
Horizontal GRF	Peak Breaking			3
	Time to Peak Breaking			1
	Peak Propulsive			2
	Time to peak Propulsive			1
Frontal GRF	Peak Medial	1		
	Peak Lateral	1		
Lower Limb Kinetics	Knee Joint Stiffness	1		1
	Ankle Joint Stiffness		1	
	Tibial Shock (Peak Positive Acceleration)	3		
	Free Moment	2		

However, it should be noted that a number of studies have failed to identify any association between vertical ground reaction force profiles and MTSS (52, 55, 151, 163, 164) (Table 8), with recent studies questioning whether ground reaction force metrics accurately represent tibial bone loading (165, 166). Matiijecich at al (165) investigated whether common vertical ground reaction force metrics such as the active vertical peak, vertical impact peak, average vertical loading rate and vertical impulse, correlated to tibial bone loads using inverse dynamics. The authors reported poor correlations between the two, concluding that ground reaction force metrics do not accurately represent the internal stress placed upon the tibia. In a further study, Loundagin et al (166) examined the fatigue behaviour of cortical bone using vertical loading rates similar to that encountered during running. Their results concluded that impact loading rate appears to have little influence on the mechanical fatigue behaviour of the bone when compared to loading cycles at lower loading rates. These results suggest that vertical ground reaction force metrics may not be as important as first thought. Considering tibial bone has been shown to have a lower fatigue life when subject to torsional loads, it is possible that the total force applied to the tibia is less important than the direction in which force is applied to the tibia. Alternatively, it may be that elevated loading rates, combined with kinematic parameters influencing torsional and bending forces, result in excessive stress applied to the medial tibia.

Creaby & Dixon (167) provide evidence to support the theory of abnormal bending forces contributing to MTSS. They reported 10 subjects with a history of MTSS to demonstrate a more medially directed ground reaction force at mid stance when compared to healthy controls. They suggested, that as the ground reaction force shifts medially, there will be an increase in the lever arm between the ground reaction force vector and the knee joint centre. As a result, there is likely to be an increase in the external bending forces at the distal tibia. Interestingly, the direction of the force vector could be influenced by frontal plane kinematics such as contralateral pelvic drop, hip adduction and rearfoot eversion (135, 136). However, Creaby & Dixon (167) failed to report kinematic data and therefore the reason for a more medially directed GRF remains unknown.

2.1.4.3 Summary and Limitations to the current literature

Medial tibial stress syndrome is a biomechanical overload syndrome of the medial boarder of the tibia, which appears to be influenced by excessive torsional, bending and shear stress. Current literature suggests the potential for several kinematic variables to influence muscular traction and torsional loading to the medial tibia, including:

- increased rearfoot eversion
- increased hip adduction

- increased contralateral pelvic drop

The current literature review also highlights evidence of an association between kinetic parameters and tibial stress fractures. Such parameters include:

- Increased vertical loading rates
- Elevated peak positive tibial accelerations
- Increased free moment
- Medially or laterally directed ground reaction force vector

It is possible that these parameters could lead to progressive overload to the bone resulting in subsequent tibial stress fracture. However, recent studies have suggested that ground reaction force variables may not accurately represent tibial bone loading. It is possible that external loads applied to the tibia only become problematic when combined with kinematic patterns that influence the direction of stress applied.

Importantly, there appears to be a lack of studies investigating proximal kinematics of the trunk and pelvis within runners with MTSS (Table 7) as well as very limited evidence reporting the kinematic characteristics of runners with current MTSS (n = 4) (55, 149-151). From a clinical perspective, understanding the kinematics associated with MTSS may allow for a greater insight into the underlying injury drivers and allow clinicians to direct appropriate treatment interventions.

2.1.5 Patellofemoral Pain Syndrome

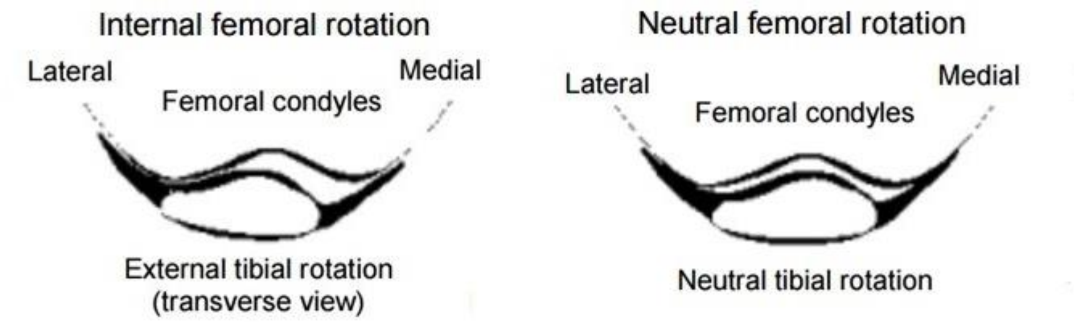
2.1.5.1 Pathomechanics of Patellofemoral Pain Syndrome

Patellofemoral pain is thought to be the result of elevated patellofemoral joint stress, resulting in increased stress to the underlying tissues including the chondral surface, subchondral bone and infrapatellar fat pad (168, 169). When exposed to repeat loading cycles, elevated joint stress may lead to an increase in chondral shear stress, an increase in subchondral bone metabolic activity (170), elevated patella bone water content (171) and excitation of nociceptors (168, 169, 172). Mechanically, elevated patellofemoral joint stress may be influenced via two mechanisms, diminished contact area between the patella and trochlea groove resulting in elevated contact pressures between the patella and femoral condyles (169), or via elevated patellofemoral joint reaction forces (168).

2.1.5.1.1 Patellofemoral contact pressures

Decreased contact area between the patella and trochlea groove may be influenced by frontal and transverse plane tibiofemoral kinematics of the femur and tibia and the resulting impact upon patellofemoral joint congruence (173). Specifically, femoral internal rotation and/ or adduction are thought to cause a medial translation of the femur underneath the patella, giving rise to elevated lateral patellofemoral contact pressures (173, 174) (Figure 6). Liao & Powers (50) investigated the effects femur and tibia kinematics upon patella cartilage stress during a 45^{o} squatting task. Systematically varying femur and tibia angles, they reported that a 6^{o} increase in femoral internal rotation resulted in a 41% increase in patella cartilage stress, while a 10^{o} increase in femoral adduction resulted in a 43% increase in patella cartilage stress (50). These results are supported by further modelling studies reporting significant increases in cartilage stress with femoral internal rotation (50, 175, 176), highlighting the influence of frontal and transverse plane kinematics of the femur on patella cartilage stress.

Figure 6: Diagrammatic illustration of tibiofemoral mechanics on patellofemoral joint positioning. Figure adapted from Lee et al (176).
Left = Internal femoral rotation and/or external tibial rotation results in a medial translation of the femur relative to the patella, reducing contact between the patella and trochlea grove and increasing contact pressures between the patella and lateral femoral condyle. Right = neutral positioning of the tibia and femur results in increased contact between patella and trochlea grove.



Several studies have also reported associations between tibial external rotation and increased patella cartilage stress (50, 175-177). It is thought that external tibial rotation shifts the tibial tuberosity lateral to the femur, creating a lateral pull on the patella leading to a subsequent increase in the contact pressures between the patella and lateral femoral condyle.

Lateral patella contact pressures may also be influenced by kinematic patterns influencing patella movement relative to the femur. In an ultrasound study by Herrington & Law (178), they found increased hip adduction angles resulted in a significant increase in lateral patella translation. They suggested that as the hip adducts, iliotibial band tension increases, resulting in lateral tracking of the patella. Although this may result from movement of the femur relative to the patella as the hip adducts, further studies have reported isolated movement of the patella to occur in response to changes to iliotibial band tension. Specifically, experimentally induced iliotibial band tension has been shown to result in significantly increased lateral patella translation, lateral patella tilt and smaller distances between the patella and lateral femoral condyle (179-181). Kinematically increased iliotibial band tension has been shown to be influenced by hip adduction (181), contralateral pelvic drop and contralateral trunk lean

(182). Therefore, it is possible that runners with PFP may exhibit either increased contralateral pelvic drop, contralateral trunk lean or increased hip adduction during running. These kinematic patterns could lead to elevated iliotibial band tension resulting in lateral tracking of the patella and elevated contact pressures between the patella and lateral femoral condyle.

2.1.5.1.2 Patellofemoral Joint Reaction Force

Lower limb kinematics may influence both the magnitude and the direction of the patellofemoral joint reaction force (PFJRF). Frontal and transverse plane kinematics can influence the mediolateral direction of the force, whereas sagittal plane kinematic may influence the posterior component of the force and therefore the compressive force acting upon the joint (168, 183). Lenhart et al (51) investigated the effects of knee flexion angle on the peak patellofemoral force, reporting that peak stance phase knee flexion angle explained up to 68% of the peak patellofemoral force. It is often thought that elevated compressive forces placed upon the patellofemoral joint, may contribute to tissue overload and pathology development (168, 184). Therefore, it is possible that runners with PFP may demonstrate increases in knee flexion angles, driving elevated patellofemoral joint stress and injury.

Based on current cadaveric and modelling studies, patellofemoral joint stress may be influenced by several kinematic patterns. Those that influence contact pressures between the lateral femoral condyle and the patella, such as increased hip adduction, hip internal rotation and tibial external rotation; and those that increase compressive forces acting on the patellofemoral joint such as increased knee flexion. The following sections will discuss the literature reporting the biomechanical characteristics of runners with PFP.

2.1.5.2 Biomechanics in Patellofemoral Pain

A visual summary of the number of studies and reported findings, investigating the difference in running kinematics and kinetics between runners with PFP and injury free controls are presented in Table 9, Table 10, Table 11 and Table 12 .

2.1.5.2.1 Kinematics

Several studies have investigated differences in hip kinematics between runners with PFP and controls including one recent systematic review with meta-analysis (185) (Table 9). Four studies reported no difference in peak hip adduction between runners with PFP and controls (186-189), while a further six case control studies (58, 190-194) and one prospective study (57) have reported runners with PFP to demonstrate significantly increased peak hip adduction when compared to controls. Combined with the pooled results from one meta-analysis (185), runners with PFP appear to demonstrate significantly increased peak hip adduction angles when compared to controls. Therefore, these findings support the pathomechanical theory that increased hip adduction could lead to greater contact pressures between the lateral femoral condyle and the patella leading to PFP (50, 169, 195).

Table 9: Visual summary of the number of studies and reported findings, investigating the difference in proximal running kinematics between runners with PFP and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. RoM = Range of Movement.

			Significantly Increased	Significantly Reduced	Non-significant
Trunk	Forward Lean	Initial Contact			
		Peak			1 (green)
	Ipsilateral Trunk Lean	Initial Contact			
		Peak			1 (green)
Pelvis	Anterior Pelvic Tilt	Initial Contact			
		Peak			1 (green)
	Contralateral Pelvic Drop	Initial Contact			
		Peak	1 (green), 1 (blue)		3 (green)
Hip	Hip Flexion	Initial Contact			
		Peak			2 (green)
	Hip Adduction	Initial Contact			
		Peak	6 (green), 1 (blue), 1 (red)		4 (green)
		RoM	1 (green)		3 (green)
	Hip Internal Rotation	Initial Contact			
		Peak	6 (green), 1 (blue)	1 (green)	7 (green), 1 (red)
		RoM	1 (green)	1 (green)	2 (green)

Current literature investigating the association between hip internal rotation and PFP remains conflicting (Table 9). Despite six case control studies reporting increased hip internal rotation amongst runners with PFP (188, 191-193, 196, 197) and one meta-analysis (185), a further seven case control studies (58, 132, 186, 187, 189, 194, 198) and one prospective study (57) have failed to identify any significant association. Although the pooled results from a recent meta-analysis suggest there is moderate evidence for a significant association between hip internal rotation and PFP (185), since publication in 2016 a further two studies have failed to identify any difference in peak hip internal rotation between runners with PFP and controls (132, 194). Therefore, whether peak hip internal rotation is commonly associated with PFP remains uncertain and should be interpreted cautiously.

The lack of conclusive evidence may possibly be explained by the poor reliability and large measurement errors associated with transverse plane kinematic measurements of the hip (199-201). During data collection of transverse plane hip kinematics, soft tissue artefact is frequently observed to cause excessive marker movement on the skin resulting in large measurement errors (202, 203). This would likely result in large between-subject movement variability as observed in many current studies and may not accurately represent true transverse plane kinematics of the femur (202, 203). For example, standard deviations as high as 7.6^{o} (194) have been reported amongst current kinematic studies, suggesting large between-subject kinematic variability. The large variability observed may subsequently limit the ability to detect small between-group differences as significant. Consequently, resulting in the failure to identify clear associations between hip internal rotation during running in PFP cohorts. Considering the strong links between hip internal rotation and patellofemoral joint stress amongst many modelling studies (50, 169, 204), the association between this parameter and PFP amongst runners may still warrant further consideration. However, there needs to be careful consideration of the reliability and validity of measurements of transverse plane hip kinematics before clinical conclusions can be made.

Knee abduction and external rotation are also thought to influence lateral patellofemoral contact pressures causing a lateral translation of the patella relative to the femur. However, to date there appears to be a limited number of studies reporting knee kinematics in runners with PFP (Table 10). One study reported increased knee external rotation amongst runners with PFP (190), while one further study has reported no difference between runners with PFP and controls (193). Two studies have reported runners with PFP to demonstrate significantly greater knee abduction angles when compared to injury free runners (187, 191), while one additional study has reported no difference (193). Therefore, it appears there is some, but limited evidence to support the association between increased knee abduction and PFP.

Table 10: Visual summary of the number of studies and reported findings, investigating the difference in distal running kinematics between runners with PFP and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. RoM = Range of Movement.

			Significantly Increased	Significantly Reduced	Non-significant
Knee	Knee Flexion	Initial Contact			
		Peak	1		5
		RoM			
	Knee Adduction	Peak	1		2
	Knee Abduction	Peak	2		1
	Knee Internal Rotation	Peak			4
	Knee External Rotation	Peak	1		1
Ankle	Ankle Dorsiflexion	Initial Contact	1		
		Peak	1		
		RoM			
Foot	Rearfoot Eversion	Initial Contact			1
		Peak			5 1 1
		Time to Peak			2
		RoM			2
		Duration			
		Velocity			3
		Toe Off			

Contralateral pelvic drop is also kinematic feature that may influence patellofemoral joint stress via several mechanisms. Contralateral pelvic drop has been shown to increase tension of the ITB (182) which may result in a lateral displacement of the patella

(180). Conversely, contralateral pelvic drop may also influence a compensatory trunk lean towards the stance limb (157) and/ or compensatory hip adduction (156, 157). Currently, only one study has reported increased contralateral pelvic drop amongst runners with PFP (58) with three further studies reporting no significant difference between PFP runners and controls (186, 187, 192) (Table 9). However, when these results were pooled within a systematic review and meta-analysis, there was found to be an increase in contralateral pelvic drop amongst runners with PFP (185). Interestingly, there is very limited evidence reporting trunk kinematics amongst runners with PFP (Table 9).

2.1.5.2.2 Kinematics influencing patellofemoral joint reaction force (PFJRF)

Peak knee flexion angles have been associated with elevated patellofemoral joint reaction forces. In a study by Lenhart et al (51) peak knee flexion during stance was found to explain up to 68% of the variance in peak patellofemoral force. Currently only a limited number of studies have reported peak knee flexion angles amongst runners with PFP (Table 10). Fox et al (191) reported increased peak knee flexion amongst runners with PFP, Bazett-Jones et al (187) reported runners with PFP to demonstrate a non-significant 3.5^{o} increase in peak stance knee flexion, while four further studies have reported no difference in peak knee flexion between runners with PFP and controls (58, 189, 194, 197). Therefore, there is currently a lack of evidence to support an association between peak knee flexion and PFP.

2.1.5.2.3 Kinetics

Elevated patellofemoral joint stress (PFJS) has been identified during walking (205) and squatting (206) amongst individuals with PFP. However, currently there is limited evidence to support an association between elevated PFJRF or peak patellofemoral joint stress amongst runners with PFP (Table 11). With respect to patellofemoral joint stress, Wirtz et al (197) reported no significant difference between runners with PFP and controls, while only one study has investigated peak PFJRF during running, reporting lower peak values amongst runners with PFP (184). However, the same study did report a more laterally directed PFJRF to be associated with PFP (184). Therefore it is possible that a more laterally directed force could contribute to elevated shear stress placed upon the patellofemoral joint during running, which may be impart explained by altered frontal and transverse plane tibiofemoral kinematics (184).

It is important to note the limitations to both modelling studies. Patellofemoral joint stress is influenced by the PFJRF and the patellofemoral contact areas to which the force is applied (PFJS = PFJRF/ contact area). With either a greater PFJRF or smaller contact area leading to increased PFJS (205). While both studies calculated the PFJRF, neither study accounted for subject specific anatomical variations which may influence contact areas (205) and therefore may not accurately represent subject specific patellofemoral joint stress. Furthermore, Wirtz et al (197) did not include transverse plane kinematics within their model. Considering transverse plane hip kinematics may influence patellofemoral contact areas, there may have been a considerable underestimation of patellofemoral joint stress within these studies.

Table 11: Visual summary of the number of studies and reported findings, investigating the difference in lower limb kinetics during running between runners with PFP and injury free controls.
Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings.

		Significantly Increased	Significantly Reduced	Non-significant
Hip Kinetics	Peak Extensor Moment			1
	Peak Abduction Moment			1
	Abduction Moment Impulse			
	Peak External Rotation Moment			
Knee Kinetics	Peak Extensor Moment			3
	Peak Abductor Moment	1		1
	Abduction Moment Impulse	1		
	Peak External Rotation Moment			1
	Peak Patellofemoral Joint Reaction Force		1	
	Peak Lateral Patellofemoral Joint Reaction Force	1		
	Peak Patellofemoral Joint Stress			1
Ankle Kinetics	Peak Plantarflexor Moment			
	Peak Inversion Moment			

There is currently limited evidence regarding the association between joint moments amongst runners with PFP (Table 11). One study reported greater internal knee abductor moment impulse in runners with PFP (207), with a further study reporting increased knee abductor moments amongst male runners with PFP (58). Conversely, additional studies have reported no difference between runners with PFP and controls for peak knee extensor moment (187, 190, 197), knee abductor moment (187), knee external rotation moment (187) or hip abductor moment (187). Similarly, there is a lack of evidence to support an association between Ground Reaction Force parameters amongst runners with PFP (Table 12). Therefore, it appears there is currently limited evidence for the role of kinetics in PFP.

Table 12: Visual summary of the number of studies and reported findings, investigating the difference in Ground Reaction Force Parameters during running between runners with PFP and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. GRF = Ground Reaction Force.

		Significantly Increased	Significantly Reduced	Non-significant
Vertical GRF	Peak Vertical Ground Reaction Force			3 1
	Vertical Impact Peak			2 1
	Time to Vertical Impact Peak		1	
	Vertical Loading Rate		1	1 1
Horizontal GRF	Peak Breaking			1
	Time to Peak Breaking			
	Peak Propulsive			1
	Time to peak Propulsive			
Frontal GRF	Peak Medial			1
	Peak Lateral			1

2.1.5.3 Summary and limitations of the current literature

The current literature suggests PFP may be influenced by kinematic patterns influencing patellofemoral contact areas resulting in elevated patellofemoral shear stress. With regards kinetic parameters, there is currently a lack of evidence to suggest kinetic parameters are associated with PFP. Conversely, substantial evidence exists to support the association between kinematic parameters and PFP. Identified parameters include:

- Increased peak contralateral pelvic drop
- Increased peak hip adduction
- Increased peak knee abduction

Several kinematic parameters including peak hip internal and knee external rotation demonstrate a plausible pathomechanical method of tissue stress, however there is conflicting evidence regarding their association with PFP. This may be due to large measurement errors and inter-subject variability associated with these parameters which therefore warrants further investigation.

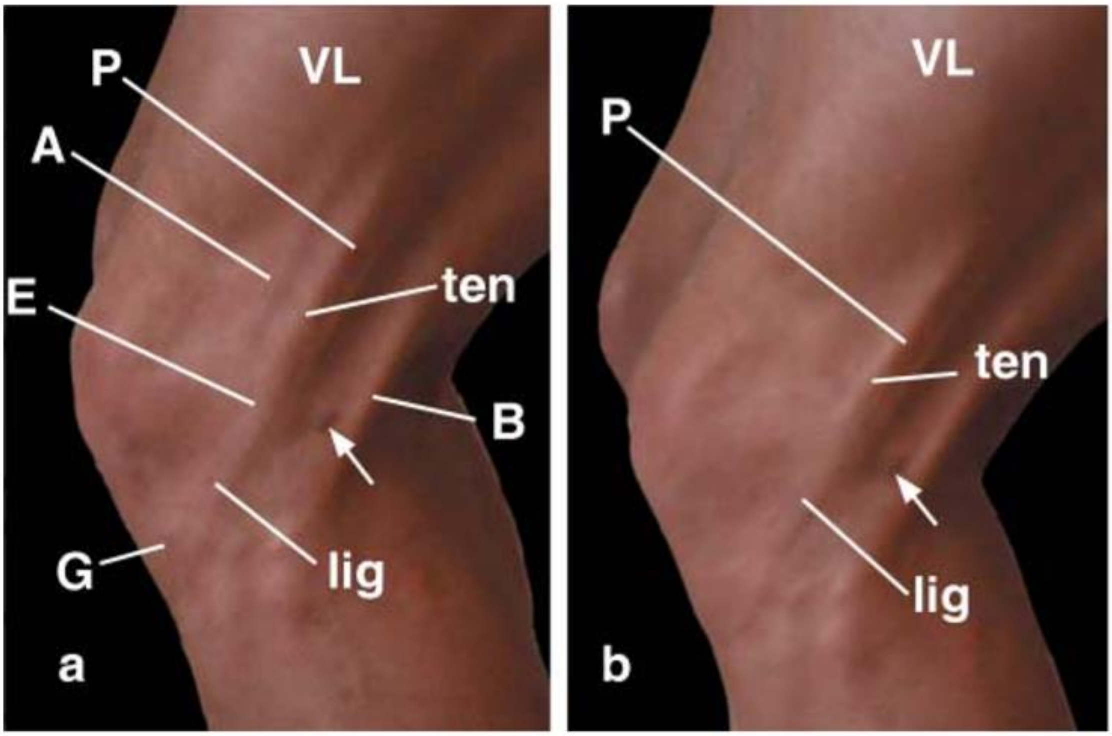
2.1.6 Iliotibial Band Syndrome

2.1.6.1 Anatomy of the iliotibial band

Anatomically the iliotibial band (ITB) is comprised of dense regular collagen and a small degree of elastin fibres (208). The ITB can be separated into both a tendinous portion and a ligamentous portion (209) (Figure 7). The tendinous portion originates from the pelvis via the gluteus maximus and tensor fascia latta, descending along the lateral thigh, continuous with the surrounding fascia, the vastus lateralis and biceps femoris (210), inserting along the linea aspera, lateral femoral condyle and lateral femoral epicondyle (208, 209, 211). The ligamentous portion arises from the lateral femoral condyle inserting into the lateral retinaculum of the patella, Gerdy's tubercle of the tiba and fibula head (209, 212). Underneath the attachment at the lateral femoral condyle lies a layer of highly vascular adipose tissue, referred to as a fat pad, containing several pressure sensitive Pacinian corpuscles (209).

The complex anatomy of the ITB means it has multiple functional roles during gait, including transmission of force from proximal muscles to the lower limbs (213), storage and return of elastic energy during gait (213) and providing anterolateral stability at the knee (214, 215).

Figure 7: Iliotibial band anatomy reprinted from Fairclough et al (209).
Figure a & b highlight the appearance of the ITB under progressively greater knee flexion angles. Ten = tendinous portion, lig = ligamentous portion, VL = vastus lateralis, B = biceps femoris tendon, E = femoral epicondyle, G = Gerdy's tubercle. As knee flexion increases between figure a & b, ITB tension can be seen to shift from the anterior portion (A) to the posterior portion (B), highlighted by the point arrow.



2.1.6.2 Pathomechanics of Iliotibial Band Syndrome

Historically iliotibial band syndrome has been considered a friction syndrome at the lateral knee (216-218). The proposed pathomechanical model was that during repeated knee flexion and extension the ITB would mechanically slip across the lateral femoral condyle (216). This repeated friction was thought to result in inflammation of an underlying bursa between the iliotibial band and lateral femoral condyle subsequently leading to pain. However, recent anatomical studies have reported that there is no such bursa between the ITB and lateral femoral condyle, and that the distal fibres of the ITB are in fact firmly anchored to the lateral femoral condyle (209, 212, 219). These findings suggest that the historic pathomechanical model of friction and bursa inflammation is not anatomically plausible (219).

More recently anatomical studies of the iliotibial band have led to suggestions that the pathomechanics are that of a compression syndrome, rather than a friction syndrome (209, 212, 219). Fairclough et al (209) conducted a series of examinations including both cadaveric examination and magnetic resonance (220) imaging of the iliotibial tract, as well as reporting the MR findings of two runners with current iliotibial band pain. The results from this study have provided several important anatomical and mechanical findings which may explain the pathomechanics of ITB pain. Firstly, their anatomical investigation identified the presence of a richly innervated and vascularised fat pad that sits in between the ITB and the femur, proximal to the lateral femoral condyle. Histologically, the fat pad itself was found to contain several pressure sensitive Pacinian corpuscles as well as myelinated and unmyelinated nerve fibres. Second, they found that as the knee moves from extension to 30^0 knee flexion, ITB tension increases and the vastus lateralis extends distally resulting in elevated compression of the fat pad. Finally, MR imaging of the runners with ITB pain found MR signal changes within the underlying fat pad. Collectively these findings suggest that mechanics influencing tension of the iliotibial band may increase compressive forces acting on the underlying pressure sensitive fat pad resulting in pain (209, 219). Considering the vast proximal and distal anatomical connections of the ITB, it is possible that ITB tension may be influenced by both proximal and distal mechanisms.

A possible distal mechanism for increased ITB tension is increased knee adduction and transverse plane tibial rotation. Cadaveric studies have reported that firm connections exist between the lateral femoral condyle and the tibia (214, 215). These connections form the ligamentous portion of the ITB acting as a stabilising structure providing resistance to knee internal rotation and adduction (214, 215). Several studies have investigated the structures of the knee responsible for providing resistance to knee internal rotation and adduction. By systematically transecting the anterolateral structures of cadaver knees, authors have found the iliotibial band provides significant resistance to knee internal rotation (221-223) and adduction (223). This suggests that the distal, ligamentous portion of the ITB acts to provide rotational stability of the knee. Therefore, it is possible that excessive knee internal rotation and adduction will increase the tension placed on the ITB and may contribute to ITBS development.

A second, possible proximal mechanism for increased ITB tension, is tissue lengthening which may be a consequence of the motion occurring at the hip and pelvis. Proximally the ITB originates from the tensor fascia latta which attaches to the pelvis, before descending to attach on the lateral condyle of the femur. Combined contralateral pelvic drop and hip adduction will lengthen the distance between these proximal and distal attachments, thereby increasing ITB tension. In support of this idea Tateuchi et al (182) used ultrasonic imaging to investigate tissue changes to the distal ITB while pelvis and hip angles were changed. They reported that hip adduction and contralateral pelvic drop resulted in a significant increase in distal ITB tension. Furthermore, in a musculoskeletal modelling study, Hamill et al (224) found ITB strain and strain rate to be significantly greater amongst runners with ITBS and that both strain and strain rate appeared to be correlated with increased hip adduction during stance. This suggests that ITB tension may be increased through tissue lengthening induced by both contralateral pelvic drop and hip adduction angles, thereby supporting the lengthening mechanism for tissue stress.

Based on the anatomy of the ITB and underlying pathology, it is possible that kinematic patterns may contribute to increased tension in the iliotibial band leading to pathology development. As such we may expect to find kinematic differences between runners with iliotibial band syndrome and healthy controls. Based on the pathomechanical model discussed, it would be likely that frontal plane pelvis and/or hip kinematics as well as frontal and transverse plane kinematics at the knee, would be expected amongst runners with ITBS.

2.1.6.3 Biomechanics in Iliotibial Band Syndrome

A visual summary of the number of studies and reported findings, investigating the difference in running kinematics and kinetics between runners with PFP and injury free controls are presented in Table 13, Table 14, Table 15 and Table 16.

2.1.6.3.1 Kinematics

Several biomechanical studies have investigated the differences in transverse plane knee kinematics between runners with ITBS and controls (Table 13). Runners with ITBS have been found to demonstrate increased peak knee internal rotation in one retrospective case control study (26, 59) and one prospective study (38) and increased peak knee adduction in two case control studies (225, 226). Based on cadaver studies of the distal ITB, these kinematics will likely increase the tension of the distal fibres of the ITB in order to resist lateral and rotational movements of the knee (214, 215, 223). However, it is worth noting that additional studies have failed to identify any difference in peak knee internal rotation (227-229) or knee adduction angles (229, 230) between runners with current ITBS and controls. Despite this, pooled results from two systematic reviews with meta-analyses suggest that based on current evidence there appears to be a significant association between increased knee internal rotaiotn and ITBS (124, 231) (Table 13).

Table 13: Visual summary of the number of studies and reported findings, investigating the difference in distal running kinematics between runners with ITBS and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. RoM = Range of Movement.

			Significantly Increased	Significantly Reduced	Non-significant
Knee	Knee Flexion	Initial Contact		1 1	2 1
		Peak		1	4
		RoM			
	Knee Adduction	Peak	2	1	1
	Knee Abduction	Peak		1	
	Knee Internal Rotation	Peak	1 1 2		2
	Knee External Rotation	Peak			2
	Tibia Internal Rotation	Initial Contact		1	1
		Peak			1
Ankle	Ankle Dorsiflexion	Initial Contact		1	1
		Peak		1	1
		RoM			1
Foot	Rearfoot Eversion	Initial Contact	1	1	1
		Peak		1	5 1
		Time to Peak			
		RoM			1
		Duration			
		Velocity			
		Toe Off			

Data to support the role of foot and tibia kinematics currently remains limited (Table 13). In one prospective study by Noehren et al (59), a small subgroup of 4 runners who developed ITBS, were found to demonstrate increased peak rearfoot eversion and tibial internal rotation. However, on a group level, no significant difference was reported when compared to controls (59). A finding further reiterated by several retrospective case-control studies (26, 228, 229, 232, 233). Interestingly, a recent meta-analysis reported runners with ITBS to demonstrate reduced peak rearfoot eversion (124), however the mechanism in which this contributes to ITBS tissue stress and injury development is poorly understood.

Currently, there is some evidence to support the theory of increased ITB tension being driven by proximal mechanics at the hip, however this evidence remains conflicting and inconclusive (Table 14). Three retrospective studies and one prospective study (59) have reported increased hip adduction in ITBS subjects (26), supporting the theory that hip adduction may increase ITB tension by increasing ITB strain associated with lengthening the proximal and distal attachment sites (224). However several further studies have either failed to identify any difference in hip adduction angles (25, 225, 227, 229, 230, 234, 235) or reported ITBS subjects to demonstrate significantly reduced hip adduction when compared to controls (228, 232).

The conflicting evidence is also compounded by the results of two systematic reviews with meta-analysis. Both Mousavi et al (124) and Aderem and Louw (231) reported that there is currently conflicting evidence suggesting no significant difference in peak hip adduction between runners with ITBS and controls. However, both meta analyses pooled the results from studies with several methodological limitations (124, 231). Furthermore, the review by Aderem et al (231) did not include the results of the prospective work by Noehren et al (59) which may have influenced their findings.

Table 14: Visual summary of the number of studies and reported findings, investigating the difference in proximal running kinematics between runners with ITBS and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. RoM = Range of Movement.

			Significantly Increased	Significantly Reduced	Non-significant
Trunk	Forward Lean	Initial Contact			
		Peak			
	Ipsilateral Trunk Lean	Initial Contact			
		Peak	1 1		1 1
Pelvis	Anterior Pelvic Tilt	Initial Contact			
		Peak	1		
	Contralateral Pelvic Drop	Initial Contact			
		Peak			2 2
Hip	Hip Flexion	Initial Contact			
		Peak			1
	Hip Adduction	Initial Contact		1	1
		Peak	1 1	2	6 2
		RoM		1	
	Hip Internal Rotation	Initial Contact			1
		Peak	1	2 1	1 1
		RoM			

There are several methodological limitations that may explain the conflicting results in the current literature. Firstly, Grau et al (228, 236) made subjects run barefoot across a 13 meter runway. Barefoot running has been suggested to cause kinematic changes including reduced hip adduction (237) and is therefore not representative of subject's normal running. Secondly, both Foch & Milner (235) and Grau et al (228) used ITBS subjects who were injury free at the time of data collection. In a later study by Foch et al (227), runners with a prior history of ITBS, asymptomatic at the time of testing, were found to demonstrate reduced hip adduction angles compared to those with current

ITBS. This suggests that kinematic patterns driving ITBS may have been resolved at the time of testing in the studies by Foch & Milner (235) and Grau et al, (228).

Finally, sex specific differences in running kinematics may also explain the conflicting results. Currently, the studies identifying greater hip adduction angles in ITBS included only female subjects (26, 59), whereas those which failed to identify differences in hip adduction included male participants only (225) or mixed sex groups (228-230, 232). Previous research has suggested female runners (238) and female ITBS runners (239), demonstrate greater hip adduction angles when compared to males. Therefore, it is possible that large methodological limitations have influenced the results of several studies and as such the association between hip adduction and ITBS remains limited.

An important kinematic consideration is the role of the pelvis and trunk, which may influence lower limb kinematics and tissue stress. Currently there is a limited number of studies investigating frontal plane pelvis and trunk kinematics in ITBS populations (Table 14). One study has reported no difference between ITBS groups and controls for trunk and pelvis kinematics (240), while a further study reported increased ipsilateral trunk lean, but no difference for contralateral pelvic drop (227). Consequently, there remains conflicting literature regarding the role of trunk lean in ITBS and a lack of evidence to support the presence of differences in frontal plane pelvis kinematics (25, 124). Reasons for the insignificant frontal plane pelvis kinematics could again be explained by the inclusion of asymptomatic subjects in the study by Foch & Milner (241). Conversely, in the study by Foch et al (227), greater ipsilateral trunk lean may have been a compensatory pattern in order to prevent excessive contralateral pelvic drop. Therefore, further research is needed to investigate the role of frontal plane pelvis and trunk kinematics in ITBS.

2.1.6.3.2 Kinetics

Currently there is limited evidence to suggest kinetics are a risk factor for ITBS. Joint moments have been studied in a limited number of studies with conflicting results (Table 15). One study reported increased internal rearfoot invertor moment to be associated with ITBS (26), whereas a prospective study reported no difference in rearfoot inversion moment between runners who developed ITBS and those who did not (59). At the knee, two studies have reported no difference in knee external rotator moment between runners with ITBS and controls (26, 59), while Stickley et al (226) reported increased external knee adductor moments in runners with ITBS. It is possible that increased knee adductor moments may increase the demand on the ITB to resist knee varus forces, however an additional study by Foch & Milner (235) reported no difference in peak knee adductor moments between runners with ITBS and controls, therefore current evidence regarding frontal plane knee moments remains conflicting. At the hip, Stickley et al (226) reported increased external hip adductor moments amongst runners with ITBS, which is in contrast to the findings of three further studies reporting no difference in frontal plane hip moments between ITBS subjects and controls (26, 59, 227). Therefore, it would appear there is limited evidence to suggest an association between runners with ITBS and joint moments.

Similarly, there appears to be limited evidence to support a link between vertical ground reaction force profiles and ITBS (Table 15). Three studies have compared ground reaction force parameters between runners with ITBS and controls (226, 229, 233). Luginick et al (229) reported greater peak vertical ground reaction force and a more medially directed ground reaction force vector amongst runners with ITBS, however two further studies reported no difference for either parameter when compared to controls (226, 233). Additionally, studies have failed to identify differences in peak horizontal breaking force (229), peak propulsive (233) and vertical loading rate (226, 233) when comparing runners with ITBS and controls. Therefore, it would appear there is limited evidence to suggest an association between ground reaction force parameters and ITBS.

Table 15: Visual summary of the number of studies and reported findings, investigating the difference in lower limb kinetics during running between runners with ITBS and injury free controls.
Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings.

		Significantly Increased	Significantly Reduced	Non-significant
Hip Kinetics	Peak Extensor Moment			
	Peak Abduction Moment	1 (green)		3 (green), 1 (red), 1 (blue)
	Abduction Moment Impulse			
	Peak External Rotation Moment			1 (green)
Knee Kinetics	Peak Extensor Moment			
	Peak Adductor Moment	1 (green)		1 (green)
	Abduction Moment Impulse			
	Peak External Rotation Moment			1 (green), 1 (red)
	Peak Iliotibial Band Strain	1 (green)		1 (red)
	Peak Iliotibial Band Strain Rate	1 (red)		
Ankle Kinetics	Peak Plantarflexor Moment			
	Peak Inversion Moment	1 (green)		1 (red)

Table 16: Visual summary of the number of studies and reported findings, investigating the difference in Ground Reaction Force parameters during running between runners with ITBS and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. GRF = Ground Reaction Force.

		Significantly Increased	Significantly Reduced	Non-significant
Vertical GRF	Peak Vertical Ground Reaction Force	1		2
	Vertical Impact Peak			1
	Time to Vertical Impact Peak			
	Vertical Loading Rate (BW/s)			2
Horizontal GRF	Peak Breaking		1	1
	Time to Peak Breaking			
	Peak Propulsive			1
	Time to peak Propulsive			
Frontal GRF	Peak Medial	1		1
	Peak Lateral			1

2.1.6.4 Summary and Limitations of the current literature

The current literature provides some limited evidence to support the associations between distal and proximal kinematics and ITBS, whereas there is a lack of evidence supporting associations with kinetic parameters. Distal mechanics may contribute to ITBS through a knee varus alignment, while proximal mechanics may influence ITB lengthening and strain. Both mechanisms could theoretically increase compression between the ITB, lateral epicondyle and underlying fat pad contributing to pathology. Kinematic patterns identified to be associated with ITBS include:

- increased peak hip adduction
- increased peak knee adduction
- increased peak knee internal rotation

It is worth noting that there is currently a lack of evidence investigating the role of proximal kinematics of the trunk and pelvis. Prior studies provide evidence to suggest contralateral pelvic drop and trunk side flexion may influence ITB tension (182, 242), however only one study has reported pelvis and trunk kinematics amongst a symptomatic ITBS population during running (227). The only other study to report frontal plane pelvis kinematics recruited runners who were injury free at the time of testing (235) and as such, may not be representative of an injured population. Therefore, further research is needed to investigate the role of frontal plane trunk and pelvis mechanics within ITBS populations.

2.1.7 Summary and Gaps

The objective of the literature review was to review and critically appraise the current literature investigating kinematic and kinetic characteristics of common running injuries in order to meet the overall aim of identifying kinematic and kinetic characteristics associated with common running related injuries (Section 1.3.7). Several databases were searched using key terms (Section 2.1.1) in order to identify research articles reporting kinematic and/or kinetic parameters associated with common running injures. These injuries included MTSS, PFP, AT and ITBS. The following sections summarise the key findings of the literature review.

2.1.7.1 Kinetics

Limited evidence was identified to support a link between kinetics and common running related injuries. Of the available evidence, only tibial stress fractures appear to demonstrate an association with kinetic parameters (52, 56, 160, 161) (Table 17). Specifically, several studies were identified reporting increased vertical loading rates (52, 56), elevated peak positive tibial accelerations (52), increased free moment (56) and a more medial directed ground reaction force vector (167) amongst runners with a prior history of tibial stress fracture. However, no consistent evidence was observed linking kinetic parameters to AT, PFP or ITBS (Table 17). This is in agreement with recent systematic reviews reporting limited evidence for an association between kinetic parameters and common running injuries (123, 128, 243, 244). Therefore, due to the lack of kinetic parameters consistently linked to common running injuries, the remainder of this thesis will focus upon kinematic associations and characteristics of common running related injuries.

2.1.7.2 Kinematics

The literature review highlighted several kinematic parameters that appear to be associated with running related pathologies. These include:

- Increased contralateral pelvic drop (MTSS & PFP)
- Increased hip adduction (MTSS, PFP, ITBS)
- Increased hip internal rotation (MTSS, PFP, AT)
- Increased knee adduction (ITBS)
- Increased knee abduction (PFP)
- Increased knee internal rotation (ITBS)
- Increased rearfoot eversion (MTSS)

Interestingly, some of the identified kinematic parameters appear to be similar across several different running injuries (Table 18). For example, increased hip adduction has been associated with PFP (58, 192), ITBS (26, 59) and MTSS (56, 152), while increased hip internal rotation has been associated with MTSS (150), PFP (188) and AT (118). Research has also suggested that due to the kinematic coupling between the femur, tibia and foot; hip adduction and hip internal rotation may influence kinematics at the rearfoot (132-134). Specifically, studies have identified correlations between hip adduction, hip internal rotation and rearfoot eversion during running and walking (132-134). Therefore, proximal hip kinematics could drive lower limb tissue stress via dynamic coupling with the foot and lower limb. This suggests that there may be several similar kinematic patterns that could underly multiple different running related injuries.

Table 17: Visual summary of the number of studies and reported findings, investigating the difference in Ground Reaction Force Parameters during running between injured runners and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. GRF = Ground Reaction Force.

		Achilles Tendinopathy			Medial Tibial Stress Syndrome			Patellofemoral Pain			Iliotibial Band Syndrome		
		Significantly Increased	Significantly Reduced	Non-significant	Significantly Increased	Significantly Reduced	Non-significant	Significantly Increased	Significantly Reduced	Non-significant	Significantly Increased	Significantly Reduced	Non-significant
Vertical GRF	Peak Vertical Ground Reaction Force			3 2			3 2			3 1	1		2
	Vertical Impact Peak			3 1			3 1 2			2 1			1
	Time to Vertical Impact Peak (s)			1			2 1		1				
	Vertical Loading Rate (BW/s)			2 1	3 2		1		1	1 1			2
Horizontal GRF	Peak Breaking			3 1			3			1		1	1
	Peak Propulsive			4 1			2			1			1
Frontal GRF	Peak Medial			1	1					1	1		1
	Peak Lateral				1					1			1

Table 18: Visual summary of the number of studies and reported findings, investigating the difference in running kinematics between runners injured populations and injury free controls. Circle colour represents the study design and the number of corresponding studies. Green = retrospective case-control study, red = prospective cohort study, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. Parameters represent peak angles.

		Achilles Tendinopathy			**Medial Tibial Stress Syndrome**			**Patellofemoral Pain**			**Iliotibial Band Syndrome**		
		Significantly Increased	**Significantly Reduced**	**Non-significant**	**Significantly Increased**	**Significantly Reduced**	**Non-significant**	**Significantly Increased**	**Significantly Reduced**	**Non-significant**	**Significantly Increased**	**Significantly Reduced**	**Non-significant**
Trunk	Forward Lean									1 (green)			
	Ipsilateral Trunk Lean									1 (green)	1 (green) 1 (blue)		1 (green) 1 (blue)
Pelvis	Contralateral Pelvic Drop				1 (green) 1 (red)			1 (green) 1 (blue)		3 (green)			2 (green) 2 (blue)
Hip	Hip Adduction			1 (green)	2 (green)		1 (green) 1 (red)	6 (green) 1 (blue) 1 (red)		4 (green)	1 (green) 1 (red)	2 (green)	6 (green) 2 (blue)
	Hip Internal Rotation			1 (green)	1 (green)		2 (green) 1 (red)	6 (green) 1 (blue)	1 (green)	7 (green) 1 (red)	1 (green)	2 (green) 1 (red)	1 (green) 1 (blue)
Knee	Adduction						2 (green) 1 (red)	1 (green)		2 (green)	2 (green)	1 (green)	1 (green)
	Abduction							2 (green)		1 (green)		1 (green)	
	Internal Rotation		1 (green)		1 (green)		2 (green) 1 (red)			4 (green)	1 (green) 1 (red) 2 (blue)		2 (green)
	External Rotation							1 (green)		1 (green)			2 (green)
Rear-foot	Eversion	1 (green)		5 (green) 2 (blue)	2 (green) 1 (red)		1 (green) 1 (red)			5 (green) 1 (red) 1 (blue)		1 (blue)	5 (green) 1 (red)

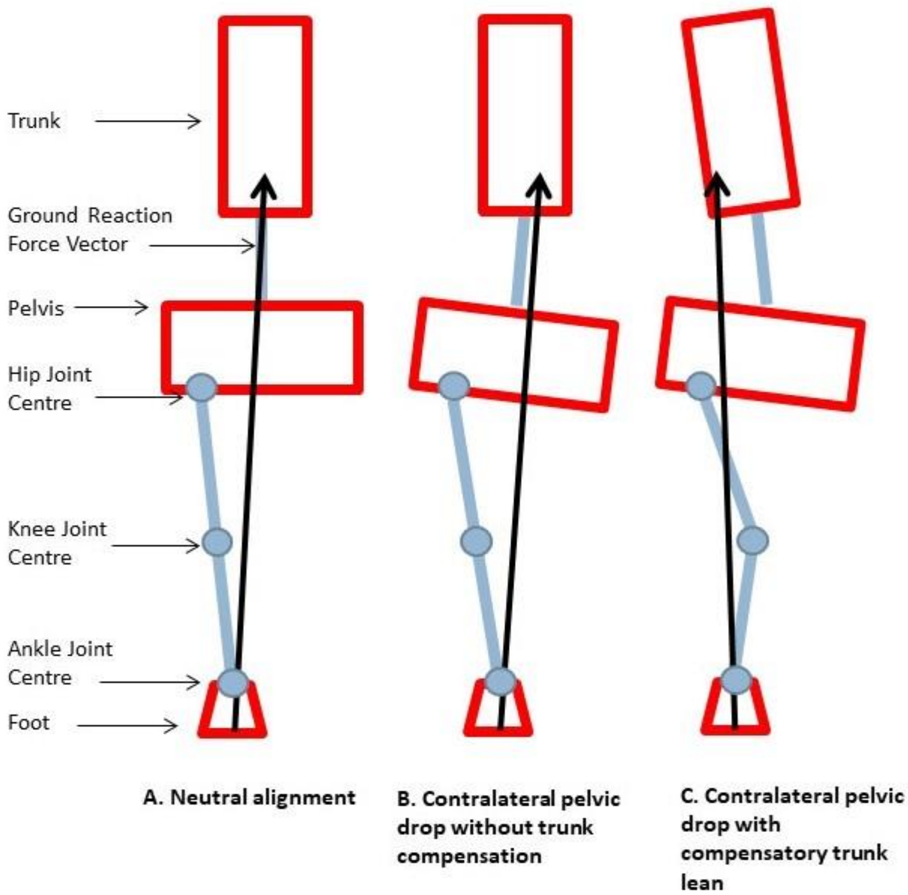
Frontal plane pelvis and trunk kinematics may also influence lower limb tissue stress and injury, however currently there is limited evidence investigating the association between the two (Table 18). In the current literature, increased contralateral pelvic drop was identified amongst runners with MTSS in only two studies (149, 150) and only one study of runners with PFP (58). No current study has investigated frontal plane pelvis kinematics in runners with AT. Interestingly, the one study reporting CPD angles amongst runners with current ITBS (227), failed to find any difference when compared to controls. However, this study did report increased ipsilateral trunk lean, which may represent a compensatory strategy to reduce peak CPD.

Only four studies were identified investigating frontal plane trunk kinematics across all injuries (187, 192, 227, 235) (Table 18). One study reported increased ipsilateral trunk lean in runners with current ITBS (227) and two studies failed to find any difference in ipsilateral trunk lean between runners with PFP and controls (187, 192). One further study reported no difference between an ITBS and control group, however the ITBS group were symptom free at the time of testing (235) and therefore may not represent an injured population. To date, no study has been identified reporting ipsilateral trunk flexion amongst runners with either MTSS or AT.

Frontal plane kinematics of the trunk and pelvis are thought to be highly coordinated and may influence lower limb tissue stress via several different mechanisms. For example, increased contralateral pelvic drop has been shown to result in a medial shift of the centre of mass, relative to the weightbearing foot (135, 136). Without compensatory trunk lean towards the standing limb, there would be an increase in the lever arm between the ground reaction force and the lower limb joint centres (Figure 8B). Through this mechanism, the knee adductor moments can increase (135), with corresponding increases in iliotibial band tension (182) and potential changes in bending forces acting on the tibia (245).

Conversely, ipsilateral trunk flexion towards the standing limb may occur as a compensation to control for excessive contralateral pelvic drop (136, 157). In such an instance, there will be a shift the force vector so it moves lateral to the knee joint centre, resulting in increased knee valgus moments. This would then drive an increase in hip adduction, knee abduction and rearfoot eversion (157, 246, 247) (Figure 8C). Through these mechanisms, it is possible that frontal plane movements of the trunk and pelvis may influence lower limb tissue stress. Consequently, further evidence is needed to investigate the association between frontal plane trunk and pelvis kinematics and common running injuries.

Figure 8: Illustrative example of the coordinated movement of the trunk & pelvis and the influence on lower limb alignment. A: Neutral trunk & pelvis allignment. B: Contralateral pelvis drop without trunk compensation. C: contralateral pelvis drop with compensatory trunk lean.



Based on the current literature review, it appears that there may be similar kinematic patterns that could underly multiple different running related injuries, in particular, kinematics at the pelvis and hip. These kinematic patterns may increase the stress placed on multiple anatomical structures which could influence injury development. Identifying such kinematic parameters would be invaluable to clinicians and researchers, as it could assist the development of both rehabilitation and injury prevention interventions.

Based on the gaps identified within the current literature, the aim of Chapter 5 of this thesis is to investigate whether similar kinematic patterns are associated with multiple different common running related injuries. Such kinematic parameters may influence the tissue load applied to multiple different musculoskeletal structures contributing to the development and persistence of common running injuries. Identifying such parameters, may subsequently assist clinicians in the development of screening and rehabilitation interventions targeted towards specific kinematic parameters. The specific research aim and objectives to be addressed are presented in Section 2.5.1.

2.2 Reliability of kinematic measurements

2.2.1 Introduction

In the previous Section (2.1.7) several kinematic parameters were identified to be associated with common running related injuries. These parameters may increase the stress placed on the musculoskeletal system during running, contributing to injury development. Interestingly, many of the kinematic parameters identified were similar across multiple different injuries. If there are common kinematic parameters associated with different running injuries, interventions that improve kinematics may reduce tissue stress and improve clinical outcomes amongst injured runners. However, in order to identify kinematic parameters associated with common running related injuries, and the effect of clinical interventions upon running kinematics, the assessment of these parameters needs to be reliable, in order to produce repeatable results.

Three-dimensional gait analysis systems are considered the gold standard for assessment of running kinematics. These systems capture the movement of markers attached to anatomical landmarks and segments which are thought to represent the motion of the underlying skeleton (248). The kinematic data gained from these measurements are often compared across populations to identify characteristics of pathological gait and/or compared within a population across a time period to monitor the response to interventions (249). However, inherent to any measurement system is a degree of error which, if not accounted for, could lead to the misinterpretation of kinematic differences as true differences, when they are in fact due to error in the measurements (249). In order to make accurate interpretations regarding between population differences in running kinematics and the effects of interventions, the reliability of kinematic measurements and the precision of measurements needs to first be established.

The objective of this section is to review the current literature in order to establish the reliability and repeatability of kinematic measurements during running. Specific reference will be made to several kinematic parameters identified in Section 2.1.7.2 to aid interpretation of the current literature and identify gaps for further research.

2.2.2 Reliability of running kinematics

A total of 7 articles were identified reporting the within and between day reliability of running kinematics (199, 201, 250-254). These studies often report good to excellent within day reliability of kinematic measurements (199, 201, 251-253). However, poor between day reliability is frequently reported for several kinematic parameters (199, 201, 251-253).

Section 2.1.7.2 of the literature review identified several kinematic parameters to be associated with multiple different running related injuries. Peak joint angles during stance phase of running have been observed amongst injured running populations, including peak hip adduction, internal rotation, knee adduction, external rotation and rearfoot eversion. Three studies in the current literature reported reliability of peak joint angles during running (199, 252, 253). These studies have shown good to excellent within day reliability of peak frontal and transverse plane hip, knee and ankle kinematics (199), (252), however poor between day reliability is frequently reported for many of these parameters (199, 252, 253). Although the high within day reliability of kinematic measurements means interpretation of between-group kinematic comparisons may yield reliable results, the poor between day reliability has implications for the interpretation of between day changes in kinematics following interventions.

Clinical interventions have recently been directed towards improvement kinematic parameters such as peak contralateral pelvic drop, hip adduction and internal rotation (69, 70, 255). Due to their associations with running related injuries, it is thought that correcting these movement patterns may result in improved clinical outcomes amongst injured runners (49, 255). Unfortunately, the poor between day reliability of these measurements means it is difficult to ascertain whether reported changes in kinematics are the result of the intervention or between day measurement error. For example, Neal et al (255) reported a 5.1^{o} and a 2.4^{o} reduction in hip internal rotation and adduction following a gait retraining intervention. However, the standard error of measurement for hip internal rotation has been reported to range between 1.1^{o} (252) and 5.9^{o} (253), and between 0.97^{o} (252) and 2.7^{o} (199) for peak hip adduction angles. Therefore, it is difficult to discern whether the effects of gait interventions are true intervention effects or simply the result of between day measurement error (256). Consequently, this could result in the interpretation of results as being “meaningful” when they are instead the result of error in measurements (256).

In order to aid the interpretation of intervention effects on kinematics, the reliability and degree of measurement error should be reported (249). This would provide an indication of the degree of change that could be expected to occur due to measurement errors and that which represents a meaningful change in kinematics. However substantial variation in kinematic measurement errors can be found between laboratories. For example, Noehren et al (252) and Stoneham et al (253), reported SEM values for peak hip internal rotation of 1.1^{o} and 5.9^{o} respectively. This is likely due to differences in testing procedures across laboratories. For example, Noehren et al (252) tested participants during treadmill running whereas Stoneham et al, (253) utilised over ground running test procedures. Therefore, for the interpretation of kinematic measurements to be meaningful, laboratories should aim to control sources of measurement error and report both the reliability and error associated with the individual testing procedures.

Section 2.1.7.2 discussed the possibility of an association between frontal plane kinematics of the pelvis and trunk and common running injuries. Between day differences in frontal plane pelvis kinematics have also been reported in several gait retraining studies (69, 252, 255). In the current literature, only two studies have reported the repeatability of frontal plane pelvis and trunk kinematics (201, 251). However, measures of repeatability and measurement errors were calculated as an average across the entire gait cycle. Therefore, the reported reliability and measurement error may not accurately represent the measurement error at specific points of the gait cycle such as peak angles or angle at initial contact. Consequently, further investigations are required in order to establish the repeatability and measurement error when assessing between day differences in discrete kinematic parameters of the trunk and pelvis.

2.2.2.1 Marker placement error

Marker placement error is considered the leading cause of between day measurement error in 3D kinematics (257-259). Three-dimensional kinematic reconstruction relies upon the modelling of joint axis using anatomical reference frames determined by the placement of markers on key anatomical landmarks. Errors in the application and reapplication of markers to anatomical landmarks is likely to result in errors in the reconstruction of joint positions and orientations (258). In a previous study by Szcerbik & Kalinowska (260), the authors reported that a 14mm marker reapplication error at the knee joint can result in kinematic errors of up to 20^{o} in frontal plane knee kinematics during walking. Therefore, accurate and reliable identification of anatomical landmarks is essential for reliable kinematic data.

One source of marker reapplication error can be introduced through inter-tester marker placement error. Della Croce et al (261) investigated the intra and inter-tester reliability in marker placement using experienced physical therapists, reporting intra-tester reliability to be significantly greater than inter-tester reliability and 3D inter-tester marker application errors ranging from 11.5mm to 24.7mm. This finding is supported by further studies reporting between day kinematic measurement errors to be greatest when compared across different examiners (257, 262), with inexperienced examiners demonstrating the lowest between day reliability measures (257). Therefore, in order to improve between day reliability of kinematics, it is important to use the same examiner with sufficient experience identifying anatomical landmarks.

2.2.2.2 Accurate modelling of skeletal movement

The poor reliability of the kinematic measurements of hip internal rotation and rearfoot eversion, may be influenced by the modelling techniques utilised. Three dimensional kinematic measurements rely on the tracking of marker clusters attached directly to the skin surface, or on top of clothing such as the shoe (203). This may introduce additional sources of error as the motion of these markers may not accurately reflect movement of the underlying skeletal segment or may be influenced by additional movement of soft tissues, termed soft tissue artefact.

Skin mounted markers attached to the thigh have been reported to create the greatest source of error compared to any other segment (203). This has implications for the interpretation of both hip and knee kinematics. Reinschmidt et al (202) investigated differences in knee joint kinematics between skin mounted markers and intracortical bone pins, reporting average errors of 21%, 70.4% and 63.6% of the total range of motion for sagittal, coronal and transverse planes respectively. The authors reported the measurement error was predominantly influenced by motion of the thigh segment, whereas motion of the tibia appeared to be more reliable with an average measurement error of less than 3^{o} (202). Such modelling inaccuracies may explain the lack of conclusive evidence to support associations between hip internal rotation and running related injuries and remains a source of error within three-dimensional measurement systems. Therefore, caution should be taken when interpreting transverse plane hip and knee angles as these may not reflect true motion of the underlying segments and may be subject to large measurement errors.

Inaccuracies in foot measurement techniques may also explain the lack of conclusive evidence supporting a link between rearfoot kinematics and running injuries. Rearfoot kinematics are often measured using marker clusters attached to the rear of the shoe (56, 149, 152). These measurement methods have been shown to demonstrate poor between day reliability (199, 253) with the validity of these measurements also questionable (129, 131). Sinclair et al (131) investigated differences in tibiocalcaneal kinematics using skin mounted and shoe mounted markers, reporting shoe mounted markers to underestimate several kinematic parameters including peak eversion, eversion range of motion and eversion velocity. However, there are limited options available for accurate kinematic modelling of the rearfoot, as running barefoot or testing runners in unfamiliar running shoes may alter the normal kinematics of the runner (237, 263).

2.2.2.3 Measures of Reliability

Accurate reporting of reliability and measurement errors is essential for interpretation of between day differences in running kinematics. Many reliability studies report statistics such as the interclass correlation coefficient (199, 252, 253) and coefficient of multiple correlation (201, 251, 254). These statistics provide a value ranging between 0, equalling no reliability, to 1, indicating perfect reliability between measurements (264, 265). Although this provides a statistical measure for interpretation of the reliability of measurements, the statistical values are of limited clinical value. This is because the statistical values do not provide estimates of the measurement precision. Therefore, it is difficult to apply these results to the clinical interpretation of between day measurement changes (264, 266).

The standard error of measurement (SEM) is one method of reporting the precision of measurements. SEM is considered an estimate of the expected variation in scores that may occur due to random error and therefore can be used to provide an estimate as to the precision of measurement (264, 266). In the repeatability literature, several studies report SEM values for kinematic data (199, 201, 253). However, the range of SEM values for the same kinematic parameters varies considerably across studies. For example, Noehren et al (252) reported the SEM of peak hip internal rotation to be 1.1^{o}, whereas Stoneham et al (253) reported a SEM of 5.9^{o}. This is possibly explained by differences in testing procedures and between day marker placement errors across laboratories, which would likely induce between laboratory differences in measurement errors. Consequently, the SEM provides a measure of precision for the individual laboratory, however, may not accurately represent the measurement error expected across different laboratories.

In such cases where the reliability of individual laboratory testing procedures is not available, use of the minimal detectable change (MDC) may aid interpretation of findings (266). The minimal detectable change provides the minimal threshold beyond the random measurement error with a 95% confidence interval. Therefore, minimal detectable change is considered to represent the degree of change representative of a true change, greater than that which could be explained by random error (266). In cases where the precision of testing measurements is not reported, the MDC may provide a measure of the minimal change required for results to be considered true intervention effects.

2.2.3 Summary and Gaps

Current literature highlights good within day reliability of kinematic measurements across the gait cycle waveform, at discrete time points and of select parameters such as peak angles (199, 201, 251-253). This means that kinematic analysis of between-group differences may lead to reliable conclusions. This is particularly important when aiming to identify the associations between running kinematics and running related injuries. However, two important points need to be considered when interpreting such results. Firstly, some kinematic measurements may not accurately represent the underlying skeletal movement, such as measures of rearfoot eversion and hip internal rotation. Second, a degree of measurement error is expected to be present in any dataset influenced by laboratory testing procedures. Therefore, an accurate understanding of the measurement errors associated with testing procedures is necessary to aid interpretation of results.

Reporting of between day measurement errors is essential to aid clinical interpretation of meaningful kinematic differences and intervention effects. Between day reliability of kinematic measurements is generally lower than that of the within day reliability with measurement error shown to vary between reliability studies (199, 201, 253). This is likely due to differences in testing procedures and measurement errors between laboratories and testers. Based on this, we would suggest that the reliability of testing procedures as well as the SEM's, are reported by individual laboratories in order to aid clinical interpretation of the accuracy of results. In the absence of individual reliability data and SEM's, the MDC may aid interpretation of whether the reported results represent meaningful change (266).

Currently no study has reported the repeatability of discrete kinematic parameters of the trunk and pelvis during treadmill running. Many gait retraining studies utilise treadmill testing procedures to investigate between day differences in discrete kinematic parameters of the trunk (267), pelvis and lower limbs (69, 70, 255). However, as there is no prior data reporting the measurement error associated with trunk and pelvis kinematics during treadmill running, it is difficult to identify whether post intervention differences represent true intervention effects or are the result of measurement error. In order to make accurate conclusions about the kinematic characteristics associated with running injuries and the effects of gait retraining interventions, there needs to be a greater understanding of the degree of change representative of true kinematic differences, rather than that which may be expected to occur due to measurement errors.

Based on the findings of the literature review, the aim of Chapter 4 is to investigate the between day repeatability, standard error of measurement and minimal detectable change of discrete kinematic parameters of the trunk, pelvis and lower limbs during treadmill running. By achieving this objective, the results will aid the interpretation of between-group differences in kinematic parameters between healthy and injured populations and the effect of gait retraining interventions.

2.3 Exposure to training load

2.3.1 Introduction

In the previous Section (2.1.7) several kinematic parameters were identified to be associated with running related injuries. It is thought that kinematics patterns such as these may increase the load or stress placed on specific tissues during each foot contact of a run (1, 28, 49). However, running kinematics alone may be insufficient to cause injury development.

According to Bertelsens's model of running injury causation (28), injury development is the result of cumulate tissue load encountered during running, exceeding tissue load capacity. The cumulative tissue load is considered the sum of the tissue specific loads experienced per stride and the frequency of load application (Figure 1B). While running kinematics may influence the tissue load experienced per stride, without an exposure to external training load the cumulative tissue load is likely to remain relatively low, and therefore unlikely to exceed tissue capacity (Figure 2B). Therefore, it is perhaps the combination of possessing kinematic patterns that increase tissue load and being exposed to external training load that influence whether tissue capacity is exceeded.

The objective of this Section is to review the current literature to identify whether training errors are associated with running injury development and whether running kinematics are influenced by the training load runners are exposed to.

2.3.2 Training load

Training load is measured in a variety of different ways, broadly categorised as internal or external workloads. Internal workloads reflect the psychophysiological stress experienced by the athlete in the context of the external exposure (268). Whereas external workloads represent the work performed by the athlete during a session or training week, commonly measured as total distance, duration or intensity. The estimation of internal workloads relies on the measurement of the internal stress experienced by the athlete, often using metrics such as heart rate variability or rating of perceived exertion, as well as a measure of the external load application, such as session duration or distance covered (268). Although high internal training loads have been linked to injury development in several team sports (269), from a running context, to date, no study has been identified that monitors the internal workloads amongst a running population (269, 270). Instead running injury research has focused upon establishing connections between external workload measurements and injury development (23, 42).

Within current running literature, several metrics have been used to quantify external training loads. Some of the most frequent metrics used include training intensity, frequency, duration and volume (23). However, there are several limitations to the use of these metrics. Firstly, training intensity represents the internal stress imposed upon a runner (268), yet it is frequently quantified using an external measure of self-reported running pace (23). Running pace will likely impose varied physiological demands upon each runner and therefore may not accurately reflect the intensity experienced by individual runners for a given pace. Second, frequency of running may not provide a true measure of external load application. For example, high frequency of running with low session duration will lead to a lower external workload compared to a lower frequency of running with longer duration. The limitations to these methods may in part explain the lack of evidence linking running intensity and frequency to injury development (23).

Weekly training volume is perhaps the easiest method used to quantify training exposure (23, 271). Measuring the total weekly miles or kilometres provides an estimate of external training load exposure, representing the cumulative load encountered over a training week. Although it could be argued that training duration is a similar measure, runners and coaches often report their level of training exposure as distance per week (41, 272), likely due to the ease of measurement through the use of global positioning system (GPS) watches. Therefore, measurement of weekly training volume appears to be a simple and quantifiable measure of external training load commonly used by runners and coaches (41, 272).

Currently, there is conflicting evidence as to whether weekly training volume influences the risk of running related injuries (23). In two prospective cohort studies, running greater than 40 miles (or 64 kilometres) per week was reported to significantly increase the risk of injury development (43, 44). A finding further supported by several cross sectional studies, reporting injured runners to be running greater miles per week when compared to uninjured runners (45, 273-276), with two of these studies reporting injured runners to have exceeded 40 miles per week (275, 276). Conversely, there is evidence to suggest greater weekly training volumes may not increase the risk of injury. Comparing the incidence of lower extremity leg pain amongst cross country runners, Reinking et al (277) found no difference between those running more or less than 40 miles per week (64km). Further studies have also reported runners with low weekly volumes to demonstrate higher injury rates than those running higher weekly training volumes (278, 279), with some suggesting higher weekly volumes may be protective against injury (280). Although there are several methodological differences between these studies, such as retrospective or prospective reporting of injuries, these results highlight that the link between training volume and injury development is not clear.

Gabbett et al (281) proposed that high workloads alone are unlikely to be the cause of injury. Instead, the rate of workload increase is likely to be the contributing factor, with sudden acute increases overwhelming the musculoskeletal system resulting in injury development (32, 281, 282). Based on this, the acute to chronic workload ratio (ACWR) has been proposed as a model explaining the how training load may influence injury development, calculated as ratio between the most recent training period (acute load) relative to the average training load over a prior duration (chronic load) (281). Using this measure several studies have shown high ACWR's are associated with injury development (269, 270), however currently the ACWR has not been applied to running related injuries.

Despite no current study reporting ACWR amongst runners, several studies have attempted to investigate whether acute week to week increases in training load are associated with an increased risk of sustaining a running related injury. Three studies reported increased injury rates following a sudden increase in weekly training volume (46, 47, 283), while one study reported no difference in injury rates when increasing weekly training volume by either 10% or 24% (48). Interestingly, in the studies by Nielsen et al (46, 47), an increased injury risk was only found when training volume was increased by more than 30% per week, with no difference in injury rates observed between those increasing by less than 10% or 10% to 30%. This raises questions as to whether training volume alone is enough to explain the development of running related injuries and does not explain why some runners can increase their training volume and remain injury free, while others cannot.

2.3.3 Interaction between training parameters and running mechanics

One possibility is that current training load measurements do not accurately reflect the cumulative tissue load encountered for a given run (30). As proposed by Bertelsen et al (28), cumulative tissue load is the combined result of tissue specific loading per stride and the frequency of load application. Current training load measurements only consider the frequency of load application measured as duration or volume. Based on this assumption, providing run duration or volume is kept constant, the external load encountered will be similar between individuals. However, factors influencing tissue load per foot contact, will result in a cumulative tissue load that may be vastly different between individuals. For example, Lenhart et al (51) reported that a 1^{o} increase in knee flexion angle during running led to a 0.21 body weight increase in patellofemoral force per step. For a given run, this would lead to a significant increase in cumulative tissue stress when compared to a runner who had 1^{o} less knee flexion. This interaction between factors influencing tissue load per stride and the frequency of load application, may explain why some runners are able to attain relatively high training loads without developing an injury, while others become injured.

2.3.4 Effect of training volume on running kinematics

It is generally thought that the gradual and progressive increase in training load, will either allow the body to tolerate higher loads, or facilitate the development of physical qualities necessary to attain high training loads while reducing the risk of injury (32, 268). Some evidence suggests this may be the case for running kinematics, with runners adapting their kinematics in response to elevated training loads. For example, Moore et al (284) found novice runners to demonstrate significantly reduced rearfoot eversion velocity following a 10 week progressive running program. Similarly, other authors have reported trained runners to demonstrate reduced hip internal rotation angles (285), less anterior pelvic tilt (286), a more flexed knee at initial contact (286), shorter stride lengths and greater stride rate (287) when compared to inexperienced runners. It has been suggested that these kinematic differences, may represent adaptations in order to reduce the risk of injury development associated with certain kinematic patterns (284).

Currently there is limited evidence investigating whether kinematic differences occur between injury free runners who regularly run different weekly training volumes. Of the available evidence, three studies reported kinematic patterns of runners separated by weekly running volume (288-290). Floria et al (290) investigated joint coordination variability between experienced and non-experienced runners, classifying runners based on their weekly running volume as being greater than 35 kilometres per week (kmpw) or less than 25kmpw. However, their results failed to identify any significant between-group differences in coordination variability.

Two further studies have reported between-group kinematic differences when comparing high and low mileage runners using principal component analysis (288, 289). Boyer et al (288) separated runners into a high mileage (>20mpw) and a low mileage groups (<15mpw), reporting low mileage runners to demonstrate greater hip external rotation and transverse plane rotation of the pelvis away from the stance leg, greater foot external rotation, greater knee adduction and less hip adduction during stance. Clermont et al (289) classified runners as high mileage and low mileage if they ran greater than 32kmpw or less than 25kmpw, reporting high mileage male runners to run with a more flexed knee throughout the stance phase, less anterior pelvic tilt, increased pelvis rotation towards the stance leg, increased hip adduction during stance and a less abducted foot position throughout the gait cycle. High mileage females were found to demonstrate less knee internal rotation during stance, greater knee flexion during swing phase and greater ankle dorsiflexion during swing when compared to low mileage females (289).

The authors of both studies suggested that the observed differences may represent kinematic adaptations to high volume training in order to reduce the risk of injury development associated with external training loads. However, there are several limitations to these studies which should be considered. Firstly, the observed differences in kinematic patterns between high and low mileage runners have not been linked to running related injuries (see Section 2.1). Therefore, conclusions suggesting the observed differences may influence injury risk should be interpreted with caution.

Secondly, in the study by Clermont et al (289), significant differences were observed in testing speeds between the high and low mileage groups. Running speed has previously been shown to influence swing phase knee flexion angles (291), anterior pelvic tilt (292) and transverse plane pelvis kinematics (292). Therefore, it is possible that differences in running speed, rather than weekly training volumes, could explain many of the observed kinematic differences.

Finally, the training volumes used to define high and low mileage runners could be considered low, when compared to literature investigating the link between injury incidence and training volumes. Previous studies have reported higher mileage running to increase the risk of sustaining injury, particularly when exceeding 40 miles per week (64km) (43, 44, 275, 276). Whereas studies reporting kinematic characteristics of high mileage runners have included a maximum weekly training volume of 33 miles (54km) per week (289). Therefore, it is possible that the mileage groups in the current studies have not exceeded a training load exposure sufficient to trigger injury development or require kinematic adaptations to prevent injury development. Therefore, further studies are required to investigate whether differences in kinematics exist when comparing runners who regularly run weekly mileages above 40 miles per week.

2.3.5 Summary and Gaps

Based on current injury causation theories, it is likely that running mechanics may only trigger injury development if there is sufficient exposure to external training loads. Similarly, increasing training load may only result in injury if an individual already possesses an intrinsic injury risk factor such as running kinematics. Considering that higher training volumes increases the frequency of load application and kinematics may increase the tissue load per foot contact, it seems plausible to expect that runners who are able to attain high weekly training volumes while remaining injury free, could either adapt aspects of their running gait, or inherently possess kinematic patterns that reduce the stress placed on the musculoskeletal system. Alternatively, due to the low training exposure and resulting low frequency of stress application, runners who only complete low weekly training volumes may demonstrate kinematic patterns similar to that associated with running related injuries yet remain injury free as they have not been exposed to an external training load sufficient to trigger injury development.

There is limited, current evidence to support the premise that runners may adapt their kinematics in response to increasing training volumes (284, 288, 289). However, there are several gaps within the current literature. Firstly, no study has specifically focused on the discrete kinematic parameters identified in the previous Section (2.1.7.2) as associated with common running related injuries, such as peak hip adduction and contralateral pelvic drop. Second, when contrasting the training volumes within kinematic studies and those associated with injury development, questions remain whether the included high mileage runners have been exposed to sufficient external training loads to trigger injury development, or require kinematic adaptations. For example, in the study by Clermont et al (289) the average weekly training volume of the high mileage runners was 54km (33 miles) per week, whereas current training exposure literature suggests that injury risk increases when exceeding 64km (40 miles) per week (43, 44, 275, 276). Therefore, it is possible that the mileage groups in the current studies have not exceeded a training exposure sufficient to trigger injury development or require kinematic adaptations to prevent injury development.

Understanding whether kinematics differ between groups of healthy runners completing different weekly training volumes, may lend insight into kinematic adaptations necessary to attain regular high volume running while remaining injury free. This may also aid our understanding as to why some runners become injured as training loads increase, while others do not. From a clinical perspective, this information may subsequently be used to inform both preventative and rehabilitative programs targeting running kinematics.

Based on the discussed literature, the aim of Chapter 6 is to explore whether kinematic parameters associated with common running injuries are associated with weekly training load exposure. By achieving this aim, this may provide a theoretical understanding as to why some runners become injured as training volume increases, while others do not. It may also enhance our understanding of whether kinematics adaptations, if any, are required to attain regular high-volume training loads while remaining injury free. This may have implications for load management amongst runners who demonstrate kinematic parameters associated with common injuries.

2.4 Gait retraining: a clinical intervention targeting running kinematics.

2.4.1 Introduction

In the previous Section, the literature review highlighted several kinematic parameters which appear to be associated with common running related injuries (2.1.7.2). It is thought that these kinematic parameters increase the load placed upon the musculoskeletal system during each foot contact of a run. When this elevated tissue load is combined with an exposure to external training load, the cumulative tissue load may cause a runner to exceed their load capacity developing injury. Based on this premise, targeting running kinematics within the rehabilitation process may reduce the load placed on the musculoskeletal system during each foot contact and subsequently the cumulative loading across an entire run. Consequently, targeting running mechanics may lead to positive clinical outcomes amongst injured runners by reducing the stress applied to injured structures and help facilitate an increase in external training load attainable.

Gait retraining has been proposed as a movement specific intervention targeting running mechanics (49, 66, 293). The aim is to provide feedback to a runner in order to teach them how to adjust running mechanics or offload injured areas when running. Once the subject has learnt the desired running technique the aim is to then reinforce the desired running mechanics. If running mechanics increase the stress on the musculoskeletal system, it seems logical that interventions which reduce this stress may offload the injured tissue, improving both function and clinical outcomes amongst injured runners and facilitating a gradual increase in training load.

There are several different methods of gait retraining identified in the current literature, including foot strike manipulation, visual feedback, step width modification and step rate modification. The objective of this Section is to review and critically appraise the literature reporting the effects of gait retraining interventions upon running kinematics and clinical outcomes amongst injured runners. Specific focus will be placed upon the kinematic parameters commonly associated with running injuries and studies investigating the effects of gait retraining on injured populations.

2.4.2 Literature Search

In order to review the current literature, electronic databases were searched in order to identify studies investigating the effects of gait retraining on running kinematics. CINHAL, MEDLINE, SportDiscus and Web of Science were searched for all years up until April 2019. Specific search terms used and are presented in Table 19. Following identification of relevant titles, abstracts were screened for relevance and full texts were then assessed against the below inclusion and exclusion criteria. References and citations of all included studies were searched to identify any additional studies which meet the inclusion/ exclusion criteria.

2.4.2.1 Inclusion

- Studies reporting kinematic outcomes in either a healthy or injured population
- Studies reporting clinical outcomes following retraining in an injured population
- Injured population diagnosed with either AT, MTSS, ITBS or PFP
- Studies using a within subjects or case control design

2.4.2.2 Exclusion

- Studies using military populations
- Studies that do not assess running
- Studies that do not report gait retraining and instead focus on other interventions such as strength training or orthotics.
- Conference abstracts

Table 19: Gait retraining literature search: key terms and boolean operators

Search Terms
Gait retraining OR retraining OR step rate OR stride rate OR cadence OR stride frequency OR step frequency OR stride length OR step width OR foot strike OR visual feedback
Biomechanics OR kinetics OR kinematics
Running OR run OR jog OR runners

2.4.3 Forefoot, Barefoot & Minimalist Running

Foot strike manipulation teaches runners to switch from a rearfoot to a forefoot contact in order to change kinetics and kinematics that are associated with injury. When changing foot strike pattern, runners are verbally cued to land on their forefoot using phrases such as "land on the ball of your foot" or "run with light footfalls", while provided with feedback to ensure they are able to adopt a forefoot contact (73, 294). Barefoot running or running in minimalist footwear has also been proposed as a method to encourage a non-rearfoot strike pattern with studies often using cues to encourage a forefoot strike contact (295-299). Therefore, this Section will discuss the kinematic effects when transitioning to barefoot running, minimalist shoes and forefoot strike.

2.4.3.1 Hip & Pelvis Kinematics

Several studies have reported frontal and transverse plane kinematics of the hip and pelvis when transitioning to forefoot running. In healthy runners three studies have reported peak contralateral pelvic drop when transitioning from rearfoot to forefoot strike running, including one systematic review with meta analyses (300) with all studies reporting no significant change (72, 301). At the hip, one study reported a significant 1.98° reduction in peak hip adduction (72) and 4.25° reduction in hip internal rotation (72), however several aditional studies have failed to idenitfy any difference in peak hip adduction (73, 267, 302) and hip internal rotation (73, 267, 302) following transition to forefoot strike running.

With regards to barefoot running or the use of minimalist running shoes, one study reported reduced hip adduction and contralateral pelvic drop at initial contact (237), however the same study did not identify any difference in peak frontal plane angles at the hip or pelvis. Further studies have also reported no difference in frontal and transverse plane hip and pelvis kinematics at initial contact, peak joint angle, joint excursion or angle at toe off when transitioning to either minimalist footwear or barefoot (296, 298, 299, 303). Therefore, it appears that transitioning to forefoot running, minimalist shoes or running barefoot may be of limited value if targeting peak frontal and transverse plane hip and pelvis kinematics.

2.4.3.2 Knee & Ankle Kinematics

The effects of forefoot or barefoot running on frontal and transverse plane kinematics of the knee and ankle has also been investigated by several studies. When transitioning to forefoot strike running one study reported reduced knee adduction at initial contact (294), however several further studies have reported no difference in frontal and transverse plane knee kinematics at initial contact, peak angle, excursion or angle at toe off with either forefoot strike, minimalist shoe or barefoot running (73, 267, 296, 298, 299, 302, 303). Therefore, it appears there is a lack of evidence to support the use of foot strike manipulation when targeting frontal plane kinematics of the knee.

Two studies have reported a significant reduction in peak rearfoot eversion when transitioning to forefoot strike running (304) and barefoot running (305). However additional studies have reported no difference in peak rearfoot eversion angle with forefoot running (73, 306) or barefoot running (296, 298), with two studies reporting increased ankle eversion range of movement during stance when transitioning to forefoot running (73, 304). Therefore, transitioning to forefoot running, minimalist shoes or running barefoot may be of limited value if targeting rearfoot kinematics.

Interestingly, evidence from one study suggests that transitioning to barefoot running may effectively target rearfoot eversion angles amongst runners with high baseline values. Morley et al (305) separated a group of 30 healthy rearfoot strike runners into three groups based on baseline eversion values; high, middle and low eversion groups.

Following transition to barefoot running they reported reductions in peak eversion angles of 5.6°, 3.6° and 0.4° for the high, middle and low groups respectively. This suggests that kinematic improvements may be dependent upon baseline values, with those demonstrating greater baseline kinematics more adaptable to change. Therefore, future studies should consider investigating the effects of gait interventions specifically targeted to those demonstrating high frontal plane angles at baseline.

The greatest biomechanical change following a transition in foot strike pattern appears to occur in the sagittal plane. Several case-series studies and systematic reviews have shown transitioning to a forefoot strike or running barefoot results in a more plantarflexed ankle (73, 89, 294-300, 306-312) and flexed knee at initial contact (75, 294, 300, 310, 312), increased ankle dorsiflexion range of movement (73, 294, 309), reduced peak knee flexion (237, 296, 303) and knee flexion range of movement (73-75, 237, 296, 297). These kinematic adaptations have been shown to contribute to significant reductions in knee joint loading and increases in ankle joint loading and may serve to offload the knee joint when running (73, 294, 313-315). However, it is important to note that Section 2.1 of the literature review failed to identify sufficient evidence to suggest sagittal plane kinematics or knee joint kinetics are associated with running related injuries.

2.4.3.3 Effects in injured populations

Six studies have investigated the clinical and biomechanical effects of transitioning to forefoot strike running amongst runners with patellofemoral pain (73, 267, 294, 316-318). Two of these studies combined cues to land forefoot with an increase in stride rate, reporting significant improvements in pain measured on a visual analogue scale (318), anterior knee pain scale (317) and knee outcome survey of activities of daily living (318) following the retraining period. The remaining four studies reported the effects of an isolated transition to forefoot strike running reporting significant reductions in pain measured on the visual analogue and numerical rating scales (267, 294), as well as improvements in the Knee Injury and Osteoarthritis Outcome Score (73), Anterior Knee Pain Scale (267), Lower Extremity Functional Scale (267), Kujala Score (316) and Patellofemoral Pain Score (316). Importantly, these improvements were above the minimally clinical important difference for the respective outcome measures (73, 267, 294, 316-318) suggesting transitioning to forefoot strike is a clinically effective rehabilitation option for patellofemoral pain.

However, despite the positive clinical outcomes, transitioning to forefoot running has several limitations. Firstly, transitioning to forefoot, barefoot or minimalist running does not address many of the kinematic patterns which may underly common running related injuries. Instead, foot strike transition appears to influence predominantly sagittal plane kinematics and kinetics at the knee and ankle, neither of which have been linked to the development of common running injuries (see Section 2.1.7). Common kinematic parameters associated with running related injuries include contralateral pelvic drop, hip adduction, internal rotation, knee abduction, adduction and rearfoot eversion, all of which have not been conclusively shown to be influenced by methods of foot strike transitioning (Table 20). Instead transitioning to forefoot running appears to shift joint loads away from the knee and to the ankle, effectively unloading the painful knee rather than addressing potential injury causing mechanics.

A second limitation is that shifting the load distribution from the knee to the ankle joint, may increase the risk of lower limb injury. Several studies have reported transitioning foot strike patterns to result in large increases in ankle plantar flexor moments, Achilles tendon loads (303, 315, 319-321) and gastrocnemius muscle activity (309). These sudden changes in load distribution may exceed the internal tissue capacity of the lower limb, resulting in injury development. In support of this idea, several studies have reported an increased incidence of lower limb pain and injuries following transition to forefoot, barefoot and minimalist running (73, 322, 323). Therefore, due to the associated injury consequences, transitioning to forefoot running may not present a safe option for the treatment of many running related injuries.

Finally, transitioning to forefoot, barefoot and minimalist shoe running may not be applicable to many runners and other injuries. For example, the increased ankle joint loading and Achilles tendon forces associated with forefoot strike running may exacerbate conditions such as Achilles tendinopathy or medial tibial stress syndrome.

This gait intervention would also not be applicable to those who already run with a forefoot strike. Instead, it may be more beneficial for gait interventions that target the specific kinematic patterns associated with common running injuries. This would offer a practical rehabilitation method that may apply to multiple different running related injuries.

2.4.3.4 Summary of Forefoot, Barefoot & Minimalist Running

The current literature highlights that forefoot, barefoot and minimalist running appears to have the greatest impact upon sagittal plane kinematics at the knee and ankle. The observed kinematic changes decrease knee joint loading and increase ankle joint loading. Although current literature suggests this may prove beneficial to reduce pain in runners with patellofemoral pain, there are several limitations to the practical and clinical applicability of foot strike transition. These limitations include increased risk of lower limb injury and the lack of evidence suggesting foot strike transition has an influence on kinematic parameters associated with common running related injuries. Therefore, gait interventions that address kinematic patterns associated with common running injuries, with low injury risk are required.

2.4.4 Visual Feedback

Visual feedback of running kinematics has been used as a method of gait retraining (69, 70). Current studies have utilised real-time feedback of 3D kinematics (70, 324) or the use of a mirror placed in front of the runner (69) as a form of visual feedback. The feedback process involves highlighting the faulty movement pattern using visual feedback combined with the use of verbal cues to facilitate changing the movement pattern. Once the runner can successfully adopt the movement pattern, a faded feedback design is utilised, whereby run duration is gradually increased and feedback gradually reduced to facilitate retention of the desired movement pattern. The following sections discuss the literature reporting the effects of visual feedback on running kinematics associated with common injuries and the effects on injured populations.

2.4.4.1 Hip & Pelvis Kinematics

Two studies have reported improvements in frontal plane pelvis and hip kinematics using visual feedback with injured runners (69, 70). Noehren et al (70) used real time visual feedback of hip adduction kinematics, while Willy et al (325) used a mirror to provide visual feedback of hip adduction. Both studies utilised a faded feedback design, where feedback is gradually reduced over 8 sessions across a two-week period. Following retraining, Noehren et al (70) reported a significant 2.3^{o} reduction in contralateral pelvic drop and 5.1^{o} reduction in hip adduction, while Willy et al (69) reported reductions of 1.9^{o} for peak contralateral pelvic drop and 5.9^{o} for hip adduction. These changes were maintained at both 1 month (70) and 3 month follow up (69), highlighting the successful use of visual feedback when targeting frontal plane hip and pelvis kinematics. Interestingly, both studies failed to identify any difference in peak hip internal rotation following the intervention.

2.4.4.2 Knee & Ankle Kinematics

Only one study has been identified reporting transverse plane knee and ankle kinematics following visual feedback (324). Hunter et al (324) provided real-time feedback of the pelvis in the transverse plane in attempt to reduce pelvis external rotation. Although pelvis rotation did not change in the direction targeted, they did report reductions in knee and ankle external rotation. However, visual feedback was combined cues to "keep the knee pointing forwards" and "keep the foot pointing forwards". Therefore, it is possible that the verbal cues used, rather than the visual feedback of the pelvis, may have resulted in the changes to knee and ankle kinematics. However, this study only included one subject and therefore further investigations are required to investigate the effects amongst larger cohorts.

2.4.4.3 Effects in injured populations

Three studies have reported improved clinical outcomes amongst injured runners following visual feedback (69, 70, 324). Hunter et al (324) reported improved VAS and LEFS scores in one runner with ITBS, however LEFS scores did not exceed the MCID of 9 points (326) and the small sample size means these results may not be generalisable to wider ITBS populations. Willy et al (69) and Noehren et al (70) both reported large reductions in pain and improvements in function to be associated with improvements in frontal plane kinematics of the hip and pelvis. Following mirror retraining, Noehren et al (70) reported a reduction in NRS from 5, to 0 out of 10 at one month follow up and a 19-point improvement in LEFS. These results are similar in magnitude to that reported by Willy et al (69), reporting a 12.1 point increase in LEFS and an NRS of less than 1 out of 10 post retraining. Importantly, the magnitude of clinical improvement exceeding the MCID in both studies, highlighting the effectiveness of visual feedback for improving both kinematic and clinical outcomes.

2.4.4.4 Summary of Visual Feedback

Current studies highlight that visual feedback of running kinematics can be used to successfully retrain frontal plane hip and pelvis kinematics (69, 70, 185) (Table 20). The observed improvements in kinematics are associated with improved clinical outcomes amongst runners with patellofemoral pain (69, 70). This highlights that interventions targeting kinematic patterns associated with running related injuries, can produce positive clinical outcomes amongst injured runners, a finding reiterated by a recent systematic review with meta-analysis (185). However, there are several practical limitations to the use of visual feedback which mean this method of retraining may be difficult to integrate into in the clinical setting. Firstly, real-time kinematic feedback requires access to 3D gait analysis technology. These systems are often expensive and are not widely available in clinical practise. Secondly, although mirror gait retraining is easy to integrate into the clinical setting, it restricts runners from continuing their normal routine and may not be a practical option for runners outside of the clinic. Finally, visual feedback studies use faded feedback designs, where feedback is gradually reduced over repeated sessions across a two-week period. This intensive retraining period may not be accessible for all runners and clinicians, which may subsequently limit the frequency of retraining sessions available. Therefore, there is a need for gait retraining methods that can be easily integrated outside of a laboratory setting while providing positive clinical and biomechanical outcomes.

2.4.5 Step rate manipulation

Step rate manipulation is the process of increasing running cadence, measured as steps per minute. In order to facilitate an increase in step rate, studies often use an audible metronome set to the desired step frequency and cue the runner to match their footsteps to the metronome. Similar to visual retraining studies, for long term retention of the new step rate, studies often use a faded feedback design where the movement pattern is initially practised with the metronome and then the metronome is gradually removed (255). The following sections will discuss the literature reporting the effects of increasing step rate on running kinematics associated with common injuries and the effects on injured populations.

2.4.5.1 Hip & Pelvis Kinematics

A limited number of studies have investigated the effects of increasing step rate on frontal plane pelvis kinematics during running (Table 20). Boyer & Derrick (301) investigated the effects of a 10% increase in step rate amongst habitual forefoot and rearfoot strikers, reporting a significant reduction in peak contralateral pelvic drop. However, the changes in joint angles were relatively small, with only a 0.9^{o} and a 0.5^{o} reduction in contralateral pelvic drop amongst habitual forefoot and rearfoot strike runners respectively. Neal et al, (255) are the only other study to report reductions in peak contralateral pelvic drop following a step rate intervention. In a group of 10 runners with patellofemoral pain, Neal et al (255) reported a 1.5^{o} reduction in contralateral pelvic drop following a 7.5% increase in step rate. However, it is important to note that these changes occurred following a 6-week retraining period and the authors did not report the repeatability of their testing procedures. Although the authors referred to a previous study reporting the standard error of measurement of kinematic data collection, measurement errors have been shown to vary between laboratory's (Section 2.2.2). Previous studies have reported the between day standard error of measurement for frontal plane pelvis kinematics of up to 1.7^{o} (201). Therefore, it is possible that the pre and post intervention differences could be the result of measurement error rather than true intervention effects.

Several studies support the use of step rate manipulation when targeting peak hip adduction angles (Table 20). Using healthy populations, studies have reported acute reductions in peak hip adduction of up to 1.8^{o} following a 10% increase in step rate (71, 301, 302, 310, 327). One study reported a statistically significant 2.9^{o} reduction in peak hip adduction at 2 week follow up (328) and a further study reported a 2.4^{o} reduction at 6 weeks (255) following a 7.5% increase in step rate. Conversely, in a group of 6 runners with patellofemoral pain, dos Santos et al (267) reported a 2.12^{o} non-significant reduction in peak hip adduction following a 10% increase in step rate. However, the small sample size of 6 subjects may have left this study underpowered to detect statistically significant differences.

Interestingly, despite increasing step rate by only 7.5%, the studies by Neal et al (255) and Willy et al (328) show greater reductions in peak hip adduction compared to the studies using a 10% increase. This may be explained by several methodological differences between the studies. Firstly, both Neal et al (255) and Willy et al (328) used a test-retest study design without reporting the repeatability of their individual testing procedures. Therefore, it is possible that their results may be influenced by between day measurement error. Secondly, it is possible that targeting gait retraining to runners with abnormal kinematics at baseline, results in greater kinematic adaptations than when targeting healthy runners. In the studies by Neal et al (255) and Willy et al (328), runners were recruited if they were either currently injured (255), or demonstrated kinetic parameters reported to be associated with injury (328). Conversely, the studies reporting lower angle changes following a 10% increase in step rate, recruited healthy runners only (71, 302, 310, 327). Therefore, it may be that specifically targeting gait retraining to those demonstrating abnormal kinematics at baseline, results in greater kinematic changes.

The effects of increasing step rate on peak hip internal rotation angles remain inconclusive. Using healthy populations studies have reported no difference in peak hip internal rotation following a 10% increase in step rate (71, 302). These findings are further supported by one study of runners with PFP reporting a non-significant 1.1^{o} reduction in peak hip internal rotation following a 10% increase in step rate (267).

Conversely, following a 7.5% increase in step rate, Neal et al (255) reported a 5.1^{o} reduction in peak hip internal rotation amongst runners with PFP. Reasons for the contrasting findings may be explained by kinematic differences between study populations. In the study by Neal et al (255) participants demonstrated baseline hip internal rotation angles of 9.1^{o}, this is significantly greater than the baseline values of 0.4^{o} reported in healthy subjects (71). It is possible that hip internal rotation angles of such magnitude are more responsive to interventions than the lower angles reported in healthy runners. Therefore, step rate retraining may effectively target hip internal rotation angles, providing these are large at baseline. Interestingly, in the study by Heiderscheit et al (71), despite no change in hip internal rotation following an increase in step rate, following a 10% decrease in step rate, there was a significant increase in hip internal rotation. Therefore, it is possible that changes in hip internal rotation may be greater in those who demonstrate either increased hip internal rotation at baseline, or a low step rate.

2.4.5.2 Knee & Ankle Kinematics

Several studies, including two systematic reviews (311, 329) have reported changes in sagittal plane knee and ankle kinematics following an increase in step rate. Studies reported increasing step rate to result in increased knee flexion at initial contact (51, 71), reduced peak knee flexion (51, 71, 255, 302, 327), reduced knee flexion excursion (302, 327), reduced ankle dorsiflexion at initial contact (310, 327, 330), lower foot inclination angle at initial contact (71, 331) and reduced peak ankle dorsiflexion (51). Therefore, there appears to be sufficient evidence supporting the use of an increase in step rate when targeting sagittal plane knee and ankle kinematics.

Limited evidence exists to support the use of step rate retraining when targeting frontal and transverse plane kinematics of the knee and ankle (Table 20). One study reported reduced peak knee abduction following a 10% increase in step rate (302), however one additional study has reported no change in frontal plane knee kinematics (267). Several further studies have reported no change in transverse plane knee kinematics following an increase in step rate (267, 301, 302, 327).

At the foot and ankle, one study reported a significant 1.6^{o} reduction in peak rearfoot eversion following a 10% increase in step rate (301). However, this was only observed amongst habitual forefoot strikers, rearfoot strikers demonstrated only a 0.2^{o} reduction in rearfoot eversion. One further study reported no significant change in peak rearfoot eversion (327). Therefore, it appears there is insufficient evidence to support the use of a step rate increase when targeting frontal and transverse plane kinematics of the knee and ankle (Table 20).

2.4.5.3 Effects in injured populations

Step rate retraining has been used as a rehabilitation intervention for patellofemoral pain in several studies. Bonacci et al (317) compared the effects of gait retraining to foot orthoses amongst 16 runners with PFP. They reported greater improvements in VAS score and Anterior Knee Pain Score amongst the gait retraining group, with 86% of the runners in the gait retraining group reporting feeling “moderately better” compared to only 29% in the orthosis group. This highlights gait retraining may be a more effective clinical intervention when compared to the use of foot orthosis. However, the gait retraining intervention consisted of an increase in step rate combined with verbal cues to land with a forefoot strike pattern. Therefore, it is not known if the clinical outcomes are the result of a step rate increase or transitioning to a forefoot contact.

Escuilier et al (318) conducted the only randomised clinical trial using gait retraining as one of the intervention arms. A total of 69 runners with patellofemoral pain were randomised into one of three intervention groups; a group receiving education on graded load management, a group receiving gait retraining combined with education on load management and a group receiving strength training combined with education. At 20 week follow up all three groups reported significant improvements in VAS scores and Knee Outcome Survey of Activities of Daily Living (KOS-ADL). However, no difference was observed when comparing between interventions. Therefore, the authors concluded that the use of gait retraining provides no additional clinical benefit when compared to education on load management. This raises questions as to the clinical effectiveness of increasing step rate when compared to alternative interventions. However, there are several limitations to this study. Firstly, gait retraining consisted of both a step rate increase and cues to adopt a non-rearfoot strike pattern. Therefore, similar to Bonacci et al (317), it is unknown if the clinical outcomes are the result of increasing step rate or transitioning to a forefoot contact. Secondly, Escuilier et al (318) did not report kinematic data, consequently it is unknown if the retraining positively impacted kinematic patterns which may drive tissue stress associated with PFP.

Only two studies have investigated the isolated effects of a step rate increase amongst runners with PFP (255, 267). Dos Santos et al (267) compared the effects of a 10% increase in step rate to foot strike transition and increasing forward trunk lean. Despite reporting improvements in LEFS, AKPS and VAS for pain, the authors reported greater improvements amongst the foot strike transition and forward trunk lean groups following the initial retraining period. Consequently, the authors questioned the effectiveness of increasing step rate when compared to alternative interventions. Questions regarding the clinical effectiveness of step rate retraining are further raised by the results of Neal et al (255). Despite reporting a 2.1- and 3.9-point reduction in average pain and worst pain measured on a numerical rating scale, at six week follow up only 3 of the 10 subjects were asymptomatic, with a further six participants reporting greater than 3/10 pain on the NRS. Pain equal to, or greater than 3 out of 10 means these subjects continue to meet the inclusion criteria for many gait retraining interventions (69, 70, 255, 317, 318). Therefore, this raises questions as to the clinical effectiveness of increasing step rate amongst runners with PFP.

One reason for the questionable clinical outcomes following step rate retraining may be explained by the lack of specificity in participant inclusion criteria. In the mirror retraining study by Willy et al (69) and the real time retraining study by Noehren et al (70), participants were only included in the retraining providing they demonstrated aberrant hip and pelvis kinematics at baseline. As such, participants reported pain scores of less than 1 out of 10 at follow up. This contrasts with step rate retraining interventions where the magnitude of baseline kinematics was not considered and participants continued to report average pain scores greater than 3 out of 10 following retraining (255, 267, 318). Considering patellofemoral pain is known to have a multifactorial aetiology, it is possible that these studies included several subjects for whom abnormal kinematics were not the underlying cause of injury. As such they would be unlikely to respond to a clinical intervention which aims to improve running kinematics. Therefore, future studies should consider targeting step rate interventions to injured populations who demonstrate abnormal kinematics at baseline. Ensuring the specificity of interventions provided may improve clinical outcomes amongst injured runners.

2.4.5.4 Summary of Step Rate Manipulation

Current studies highlight that increasing step rate can successfully reduce kinematic patterns associated with running related injuries. Specifically, increasing step rate has been shown to reduce peak hip adduction and peak contralateral pelvic drop (Table 20) which have commonly been associated with several running injuries (Table 18). Although the evidence for reducing hip internal rotation remains limited, it appears that increasing step rate may successfully reduce hip internal rotation angles in participants who demonstrate either low step rate at baseline or large hip internal rotation angles. Therefore, step rate retraining is an appropriate intervention when aiming to improve kinematic patterns commonly associated with running related injuries.

Unfortunately, the clinical effectiveness of increasing step rate amongst injured populations remains inconclusive. Although reductions in pain have been observed, the magnitude of clinical improvement appears to be less than that observed following visual feedback (69, 70). One reason for this may be the lack of specificity in participants targeted with gait retraining. In contrast to visual gait retraining, no current study has specifically targeted an increase in step rate to injured runners who demonstrate abnormal hip and pelvis kinematics at baseline. By targeting interventions to the underlying mechanics there may be a greater improvement in clinical outcomes observed. Therefore, future research is needed to investigate whether a step rate increase targeted at injured runners who demonstrate abnormal kinematics at baseline can enhance clinical and functional outcomes.

2.4.6 Step width manipulation

Step width manipulation is the process of increasing or decreasing a runner's step width in order to influence a mechanical change. Systematic variations in running step width can be facilitated by placing a taped line along the centre of a runway or treadmill and verbally cueing runners to land either side of the line (245, 332). The following sections will discuss the effect of increasing running step width upon kinematic parameters associated with common running related injuries.

2.4.6.1 Hip & Pelvis Kinematics

No study has reported kinematics of the pelvis when running with different step widths, whereas two studies have reported hip kinematics following cues instructing participants to run with a wide, narrow and preferred step width (68, 333) (Table 20). Only one study has reported peak hip internal rotation angles, however it did not observe any change across step width conditions (333). Two studies have reported reduced peak hip adduction angles when running with a wide step width compared to narrow and preferred step widths (68, 333). Therefore, cueing runners to run with a wider step width seems an effective retraining method to reduce peak hip adduction angles.

2.4.6.2 Knee & Ankle Kinematics

Two studies have reported the effects of step width on transverse plane knee kinematics (68, 333), however no study has reported the effects upon frontal plane kinematics. Peak knee internal rotation has shown to significantly reduce when running with a wide step width, but only when compared to narrow step width conditions (68, 333). Therefore, when targeting peak knee internal rotation angle, increasing step width is likely to be effective only if the individual demonstrates a narrow step width at baseline.

Peak rearfoot eversion angles have also been reported in two studies. Brindle et al (333) reported a 1.1^{o} and 1.5^{o} reduction in peak rearfoot eversion when increasing step width from narrow to preferred and from preferred to wide. Therefore, suggesting step width manipulation could effectively reduce peak rearfoot eversion. However, Pohl et al (332) reported rearfoot eversion and rearfoot eversion excursion to only be reduced when increasing step width from a narrow condition. Therefore, when targeting peak rearfoot eversion, or rearfoot eversion excursion, it is possible that increasing step width is only effective if the individual demonstrates a narrow step width at baseline.

2.4.6.3 Effects in injured populations

Currently no study has investigated the effects of increasing step width on injured populations. Therefore, despite some evidence to suggest increasing step width may target kinematic parameters associated with running injuries, it is unknown if the magnitude in change would be sufficient to provide positive clinical outcomes.

2.4.6.4 Summary of Step Width Manipulation

Currently there is limited evidence investigating the effects of step width retraining on running kinematics. Based on the available literature, there is some evidence to support the use of step width manipulation when targeting peak hip adduction, however knee internal rotation angle and rearfoot eversion may only be effectively targeted in runners with a narrow step width at baseline. Therefore, further investigations are needed to first support the effects of step width manipulation on kinematic parameters and whether this can successfully influence clinical outcomes amongst injured populations.

2.4.7 Summary and Gaps

In Section 2.1.7.2 the literature review highlighted several kinematic parameters associated with running related injuries. These include parameters such as peak contralateral pelvic drop, hip adduction, hip internal rotation, knee adduction/ abduction, knee internal/ external rotation and rearfoot eversion. The purpose of this Section was to review the gait retraining literature to establish whether methods of gait retraining can effectively target these kinematics parameters and improve clinical outcomes amongst injured runners.

Several methods of gait retraining were reviewed including foot strike transition and barefoot running, visual feedback of running kinematics, increasing step rate and increasing step width. Of these methods visual feedback and step rate increase appear to effectively reduce frontal plane pelvis and hip kinematics amongst both healthy and injured running populations. However, there is limited evidence to suggest there is sufficient impact upon frontal and transverse plane knee and ankle kinematics. Evidence from two studies appears to suggest increasing step width may reduce hip adduction angles as well as rearfoot eversion and knee internal rotation (68, 333), but the latter are likely to only be influenced in those with a narrow step width at baseline. Foot strike transition and minimalist shoes do not appear to have a significant influence on kinematic parameters commonly associated with running related injuries. As such, foot strike transition may not be an intervention of choice if the aim is to improve kinematic parameters associated with running related injuries. Based on the current available literature, step rate retraining and visual feedback appear to be effective interventions when targeting frontal plane pelvis and hip kinematics, while step width retraining may influence rearfoot eversion and knee internal rotation in participants with a narrow step width.

Visual feedback, step rate retraining and transitioning foot strike are the only gait retraining interventions to report clinical outcomes amongst injured populations. Transitioning to forefoot running has been shown to produce positive clinical outcomes in runners with patellofemoral pain, however as previously discussed (Section 2.4.3.3), there are several limitations to this method of gait retraining reducing its clinical value. Specifically, transitioning to forefoot running does not appear to impact the kinematic parameters commonly associated with running related injuries (Table 20) and has been shown to increase the risk of lower limb injury development. Visual feedback has been shown to successfully target hip and pelvis kinematics and improve clinical outcomes amongst injured runners. However retraining methods are predominantly laboratory based, requiring close clinical monitoring. Such methods may not be feasible in clinical practise and as such there is a need for gait retraining interventions that can be easily integrated outside of the laboratory and into a runner’s normal routine.

Table 20: Visual summary of the number of studies and reported findings, investigating the effects of gait retraining interventions kinematics parameters associated with common running injuries. Circle colour represents the study design and the number of corresponding studies. Green = case-series study of injury free runners, orange = case-series study of injured runners, blue = meta-analysis findings of a systematic review. Gaps indicate no reported findings. Foot strike manipulation includes studies reporting either transition to barefoot or minimalist shoes and those directly curing a fore foot contact. Visual feedback includes studies utilising mirror or 3D real time feedback.

			Step Rate Increase		**Foot Strike Manipulation**		**Step Width Increase**		**Visual Feedback**	
			Significantly Reduced	**Non-significant**	**Significantly Reduced**	**Non-significant**	**Significantly Reduced**	**Non-significant**	**Significantly Reduced**	**Non-significant**
Trunk	Ipsilateral Trunk Lean	*Peak*								
Pelvis	Contralateral Pelvic Drop	*Peak*	orange 1, green 1			green 3, blue 1			orange 2	
Hip	Hip Adduction	*Peak*	green 6, orange 1	orange 1	green 1	green 6, orange 2, blue 1	green 2		orange 2, blue 1	
Hip	Hip Internal Rotation	*Peak*	orange 1	orange 1, green 2	green 1	green 4, orange 2		green 1		orange 2
Knee	Adduction	*Peak*		orange 1		green 4, orange 2				
Knee	Abduction	*Peak*	green 1		green 1	orange 1, green 2				
Knee	Internal Rotation	*Peak*		green 2		green 1, orange 1	green 2			
Knee	External Rotation	*Peak*		green 1, orange 1		green 4, orange 1	green 1		orange 1	
Rearfoot	Eversion	*Peak*	green 1	green 1	green 2	green 3, orange 1	green 1			

Step rate retraining is one method of retraining that could be integrated outside of the laboratory (328). Through the use of Global Positioning System (GPS) "smart" watches and portable mobile metronome applications, runners may be able to self-retrain and monitor their step rate without the need for close clinical supervision (328, 334). However, despite reporting reductions in pain amongst injured populations, average pain scores have remained greater than 3 out of 10 following step rate retraining (255, 267, 318). This contrasts with visual retraining methods reporting pain scores of less than 1 out of 10 following retraining (69, 70). Therefore, the clinical effectiveness of step rate retraining when compared to other interventions remains questionable (267, 318).

In order to improve the efficiency of gait retraining it may be necessary to ensure interventions are specifically targeted to those who demonstrate abnormal running kinematics at baseline. Considering many running related injuries are known to have multifactorial aetiologies, it is possible that interventions must be targeted to the appropriate injury contributors in order to improve clinical outcomes. In the current literature Noehren et al (70) and Willy et al (69) are the only studies to specifically target gait retraining to runners with abnormal running kinematics at baseline. Consequently, the improvements in clinical outcomes were much greater than that of step rate retraining studies. Therefore, future research is required to investigate whether step rate retraining, targeted at those demonstrating abnormal hip and pelvis kinematics at baseline, improves clinical outcomes amongst injured runners.

Based on the findings of the current literature review the aim of Chapter 7 is to investigate whether a simple method of gait retraining can be used to effectively improve biomechanics and improve clinical and functional outcomes amongst injured runners. The specific objectives were to investigate whether a 10% increase in running step rate influences frontal plane kinematics of the hip and pelvis, as well as clinical outcomes in runners with PFP. Secondary objectives were to investigate whether runners can self-administer a 10% increase in step rate using an audible metronome and a GPS smart watch and whether these changes can be maintained at short term and long term follow up.

The impact of achieving this aim is to provide preliminary evidence for the clinical effectiveness of a simple method of gait retraining amongst runners with PFP. The method employed can be easily integrated into clinical practise without the need for close clinical supervision and therefore offers a practical retraining method for clinicians. Based on the premise that running kinematics increase tissue load per foot contact and training load exposure influences cumulative tissue load, interventions which target kinematic parameters associated with common running injuries may reduce pain and allow runners to increase their training load exposure.

2.5 Literature Review: Summary, Aims, Objectives, Hypotheses & Impact

The aim of this narrative literature review aim was to explore the literature to identify the kinematic and kinetic characteristics of common running related injuries, the reliability of kinematic assessment measures, whether training load exposure influences injury risk and running kinematics and whether gait retraining interventions can effectively target running kinematics. Through this review, gaps within the current literature have been identified informing specific research objectives which will be addressed within subsequent chapters of this thesis. The following section summarises the knowledge gaps identified in the preceding sections.

Within the current literature review, several kinematic parameters were identified to be associated with common running injuries. Interestingly, many of these observed patterns were similar across multiple different running injuries, suggesting that similar kinematic patterns may be associated with multiple different running related injuries. In particular, peak contralateral pelvic drop, hip adduction and hip internal rotation. It is possible that these kinematic patterns may increase tissues stress at multiple different anatomical locations leading to injuries at different sites. However, no current study has investigated whether similar kinematic parameters are associated with multiple different running related injuries. Furthermore, there is limited evidence investigating whether frontal plane trunk and pelvis kinematics are associated with common running injuries. If this is the case, rehabilitation interventions that effectively target these parameters may reduce stress placed on the musculoskeletal system contributing to improved clinical outcomes amongst injured runners.

In order to identify kinematic parameters which are associated with running injuries and the effect of clinical interventions, assessment measures need to be repeatable. Several studies have reported low between day repeatability and large measurement errors associated with frontal and transverse plane kinematics of the hip and knee. Such measurement errors are likely to vary between laboratories and testing procedures. No previous study has reported the repeatability or measurement error associated with trunk and pelvis kinematics during treadmill running, yet investigations frequently utilise treadmill testing procedures to target interventions at these parameters. As such it is difficult to establish whether between day differences in running kinematics represent true intervention effects or are the result of measurement error. Therefore, as well as establishing the repeatability of individual testing procedures, there is a need to establish the repeatability of discrete kinematics of the trunk and pelvis during treadmill running.

An interesting finding of this literature review is the limited literature investigating the interaction between training load exposure and running kinematics. It is thought that running kinematics may increase the stress placed on the musculoskeletal system during each foot contact of a run. When combined with an external training load exposure, the cumulative tissue stress placed on the musculoskeletal system may cause some runners to exceed their injury threshold at much lower training volumes than others. Therefore, understanding whether kinematics are influenced by exposure to training loads, may have implications for both research and clinical practise.

Currently, no study has investigated whether kinematic parameters commonly associated with running injuries differ between those who run high and low weekly training volumes. From a clinical perspective, differences in running kinematics may aid our understanding of why some runners become injured as training volume increases while others do not. As such, clinical interventions which target running kinematics, may be utilised within the rehabilitation process in order to allow runners to increase their training volume by reducing the stress placed on the musculoskeletal system per foot contact. Therefore, there is a need to further explore possible interactions between training load exposure and running kinematics.

Several studies provide evidence to suggest gait retraining may improve running kinematics commonly associated with running injuries and clinical outcomes amongst injured runners. However, many of the interventions require close clinical supervision and restrict runners from continuing their normal routine. Therefore, there is a need for clinical interventions that can be easily integrated outside of the laboratory and clinical setting. Increasing step rate is one gait retraining method that could be integrated outside of the clinical setting. However current intervention studies have reported participants to have continued symptoms after the intervention period, raising questions as to its clinical effectiveness. Based on current literature, it is possible that clinical outcomes following gait retraining may be improved by targeting runners with aberrant running kinematics at baseline. However, no current study has investigated whether specifically targeting step rate retraining to injured runners with sub-optimal baseline kinematics, results in greater clinical outcomes and can be integrated outside of the clinical setting.

2.5.1 Aims, Objectives, Hypothesis & Impact

Based on the identified gaps within the current literature, the following aims, objectives, hypotheses and impact were established:

1. Establish the repeatability of discrete kinematic parameters during running in order to aid the interpretation of between-group and post intervention kinematic differences. The specific objective was:
    a. To investigate the between day repeatability, standard error of measurement and minimal detectable change of discrete kinematic parameters of the trunk, pelvis and lower limbs during treadmill running (Chapter 4).

       Impact: identifying the repeatability, standard error of measurement and minimal detectable change of discrete kinematic parameters will provide a clinically meaningful threshold for which interpretation of true between day differences can be made.

2. To investigate whether similar kinematic parameters are associated with multiple different common running related injuries (Chapter 5). The specific objectives were:
    a. To investigate whether there are differences in running kinematics between a group of runners with common running injuries (ITBS, PFP, MTSS and AT) compared to a healthy control group.
    H_1: Injured runners will demonstrate greater contralateral pelvic drop, hip adduction and rearfoot eversion when compared to controls.
    b. To investigate whether kinematic differences observed between injured and healthy runners, differ between injury subgroups.
    H_0: There will be no difference in kinematic parameters between injury subgroups.

Impact: identifying kinematic parameters associated with common running related injuries may assist clinical understanding of key parameters which may increase tissue loading. Subsequently this information may aid the development of screening and rehabilitation interventions, specifically targeted towards these kinematic parameters.

3. To explore whether kinematic parameters associated with common running injuries are associated with weekly training load exposure (Chapter, 6). The specific objective was:
    a. To investigate whether there is a difference, between groups of high and low-mileage runners, in the proportion of individuals who demonstrate kinematic patterns associated with injury.
    H_1: when compared to low-mileage runners, injury-free high-mileage runners will demonstrate a lower frequency of kinematic patterns similar to those associated with common running injuries.

Impact: this may provide a theoretical understanding as to why some runners become injured as training volume increases, while others do not. It may also enhance our understanding of whether kinematics adaptations, if any, are required to attain regular high-volume training loads while remaining injury free. The possible implications for load management amongst runners who demonstrate kinematic parameters associated with common injuries are discussed.

4. To investigate whether a simple method of gait retraining can be used to improve biomechanics, clinical and functional outcomes amongst injured runners. The specific objectives were:
    a. To investigate whether a 10% increase in running step rate influences frontal plane kinematics of the hip and pelvis, as well as clinical outcomes in runners with PFP.
       H_1: a 10% increase in step rate will result in significant reductions in frontal plane hip and pelvis kinematics, improvements in clinical outcomes and function.
    b. To investigate whether runners can self-administer a 10% increase in step rate using an audible metronome and a GPS smart watch and whether these changes can be maintained at short term and long term follow up.
       H_1: runners will increase their step rate by 10% at short term follow up which will be maintained at long term follow up.

Impact: the impact of achieving this objective is to provide preliminary evidence for the clinical effectiveness of a simple method of gait retraining amongst runners with PFP. The method employed can be easily integrated into clinical practise without the need for close clinical supervision and therefore offers a practical retraining method for clinicians.

3 Chapter 3: Methods

3.1 Overview

The following section outlines participant recruitment and kinematic data collection methods used to meet the thesis aims presented in Section 2.5.1. Statistical testing procedures are presented in the appropriate section of each Chapter.

3.1.1 Ethics

Ethical approval for the research protocol was obtained via the University of Salford local ethics committee prior to the commencement of data collection (HSCR13-17a, HSCR13-17b, HSCR16-49, Appendix A). Data collection commenced in June 2013 with the initial aim of forming a kinematic database of healthy control subjects. This application was later amended to include the investigation of injured runners in line with the aims generated for this thesis (HSCR13-17b, appendix A). As the literature review progressed and research aims were developed, a further ethical application was submitted in order to investigate the effects of gait retraining upon injured runners (HSCR16-49, Appendix A). This latter trial was registered as a clinical trial via ClinicalTrials.gov (registration No. NCT03067545).

Prior to participation all participants were provided an information sheet detailing data collection protocols (Appendix B), this was discussed with the participant in detail and they were provided the opportunity to ask any necessary questions. All participants were then required to provide written informed consent prior to commencing data collection procedures (Appendix C). All data was collected in accordance with a standardised protocol developed for the purpose of meeting the aims of this thesis, as well as wider research aims within the department for use of data collected from running clinic services operated by the University of Salford. These included investigating the effects of running speed on kinematics, differences in kinematics between elite and recreational runners and the relationship between clinical assessment measures and kinematics.

3.2 Participant Recruitment

In order to meet the research aims a convenience sample of participants was recruited through a University based running clinic as well as externally, via poster advertisements located at sports injury clinics, running clubs and running race events across the Greater Manchester area. Written and verbal consent to participate was gained from all subjects prior to being included within the corresponding studies (Appendix C). All participants were screened by a qualified physiotherapist [CB] to confirm injury status, diagnosis and their eligibility prior to inclusion within the specific study protocols. The inclusion and exclusion criteria for groups is outlined in the following sections and can also be found in Appendix D.

3.2.1 Healthy Control Subjects

3.2.1.1 Inclusion Criteria

Healthy controls were defined as endurance runners competing at distances greater than 1500m, running a minimum of twice per week and having reported no injury or time off from training in the last 18 months.

3.2.1.2 Exclusion Criteria

Subjects were excluded from the control group if they reported any current injury or injury within the last 18 months. Injury was defined in accordance with a consensus definition outlined by Yamato et al (335), specifically an injury was defined as any musculoskeletal ailment causing a stoppage or restriction to running volume, duration or speed for a minimum of 7 days or three consecutive scheduled training sessions, or that required the runner to consult a physician or health care professional. Any prior history of a common running related injury, specifically including medial tibial stress syndrome, Achilles tendinopathy, patellofemoral pain or iliotibial band syndrome resulted in exclusion from the control group. Additional exclusion criteria included having only just started running in the last 2 years, any prior musculoskeletal surgery, neurological impairment, diagnosed knee or lower limb osteoarthritis or any other injury following either trauma or sporting activity.

3.2.2 Injury Groups

3.2.2.1 Inclusion Criteria

Injured subjects were recruited if they were running a minimum of twice per week and reported a limitation to their running volume due to a running related injury. Pain experienced must have been of an insidious or gradual onset during running, rated a minimum of 3 out of 10 on a numerical rating scale (NRS) for worst pain experienced in the past week (0 = no pain, 10 = worst possible pain) and meet the consensus definition of a running related injury outlined by Yamato et al (335). Specifically, an injury was defined as any musculoskeletal ailment causing a stoppage or restriction to run volume, duration or speed for a minimum of 7 days or three consecutive scheduled training sessions, or that required the runner to consult a physician or health care professional. Injuries were assessed and diagnosis confirmed by the lead clinician and researcher [CB]. Specific injury diagnostic criteria are outlined below.

3.2.2.2 Exclusion Criteria

Subjects were excluded if the onset of injury was due to trauma or any other sporting activity. If they have had prior traumatic injury to the area, if there is any history of surgery or pre-existing medical conditions that may affect their gait or if there were any abnormal findings on a standardised physical assessment that may have influenced their gait or symptoms (Appendix D). To control for training errors as a potential underlying cause of injury, participants were also excluded if they reported the onset of symptoms to occur following an increase in their weekly training volume equal to, or greater than 30% (47). Participants were also excluded if they reported having less than 2 years running experience.

3.2.3 Patellofemoral Pain Syndrome

The diagnosis of patellofemoral pain syndrome was made in accordance with previously published diagnostic criteria (336) and previous biomechanical studies (58, 294). Specifically, for inclusion to the study participants must report a subjective history of retropatella or peripatellar knee pain, reproduced on squatting and any one or more of the following: stair ascent/ descent, kneeling, prolonged sitting, hopping or jumping.

Pain on squatting has been shown to have a sensitivity of 91% and a negative predictive value of 74% suggesting this test to be the best available test for PFP (337, 338). A combination of additional, but non-essential, clinical tests was used to further increase the diagnostic accuracy of PFP (339). Tests included patella compression, patella apprehension, pain on palpation of the lateral patella facet and pain on resisted quadricep contraction in 30^{0} knee flexion (338, 339).

Participants were excluded if they presented with any signs of patella instability, ligamentous or meniscal injury (336) determined following a standardised physical examination procedure including McMurrys grind test, Apleys grind test, Lachmans, anterior draw, posterior draw, Nobles test, varus and valgus stress testing.

3.2.4 Iliotibial Band Syndrome

The diagnosis of ITBS was made in accordance with the diagnostic criteria outlined in previous biomechanical studies (228) and also described by Fredericson (217). Specifically, pain must have been of insidious onset presenting as a sharp or burning pain localised to the lateral femoral condyle that is aggravated and may worsen during running, easing with cessation of running. Additional subjective symptoms include pain aggravated by downhill running and stair decent (217). On objective examination subjects must have presented with localised tenderness and pain reproduced on palpation of the distal ITB at the lateral femoral condyle approximately 3 cm above the lateral knee joint line and a positive Nobles compression test.

Participants were excluded if they presented with any signs of patella instability, ligamentous or meniscal injury determined following a standardised physical examination procedure. This included McMurrys grind test, Apleys grind test, Lachmans, anterior draw, posterior draw, varus and valgus stress testing, patella compression, patella apprehension, pain on palpation of the lateral patella facet and pain on resisted quadricep contraction in 30^{0} knee flexion.

3.2.5 Medial tibial stress syndrome

The diagnosis of MTSS was made in accordance with the diagnostic criteria outlined by previous authors (139, 144, 340). On subjective assessment pain must be reported to be localised to the distal third of the medial tibia exacerbated by activity and easing with rest. Participants were excluded if they presented with symptoms representative of alternative pathologies, including chronic exertional compartment syndrome or tibial stress fracture. Specifically, participants were asked if they experience pain at rest or at night, symptoms of cramping, burning, pins and needles, temperature changes in the feet or pressure/ swelling sensations exacerbated during activity (139). In the instance that these symptoms were present, participants were excluded from the study.

On objective examination pain was reproduced on the shin palpation test; pain on palpation of the medial ridge of the tibia along a 5 centimetre length (139). Participants were excluded if they had a positive shin oedema test, pain on compression of the tibial body or pain in a localised region less than 5 centimetres along the tibial boarder. These participants were excluded as findings may be indicative of bone stress injury rather than a biomechanical overload to the soft tissue structures of the medial tibial border (340, 341).

3.2.6 Achilles tendinopathy

The diagnosis of Achilles Tendinopathy was made in accordance with the diagnostic criteria outlined by Silbernagel & Crossley (342) and Hutchinson et al (343). Participants were included if pain was reported to be of gradual or insidious onset during or following running, morning stiffness that eases with movement and pain on loading activities of running or hopping that may or may not ease into activity.

On objective examination pain was required to be localised to the mid portion of the Achilles tendon, reproduced on palpation of the mid portion of the Achilles tendon approximately 2 to 6cm proximal to the insertion at the calcaneus. A combination of self-reported pain localised to the mid portion of the tendon, symptoms of morning stiffness and pain on palpation of the tendon has been shown to have a sensitivity of 83% and specificity of 89% for mid portion Achilles tendinopathy (343).

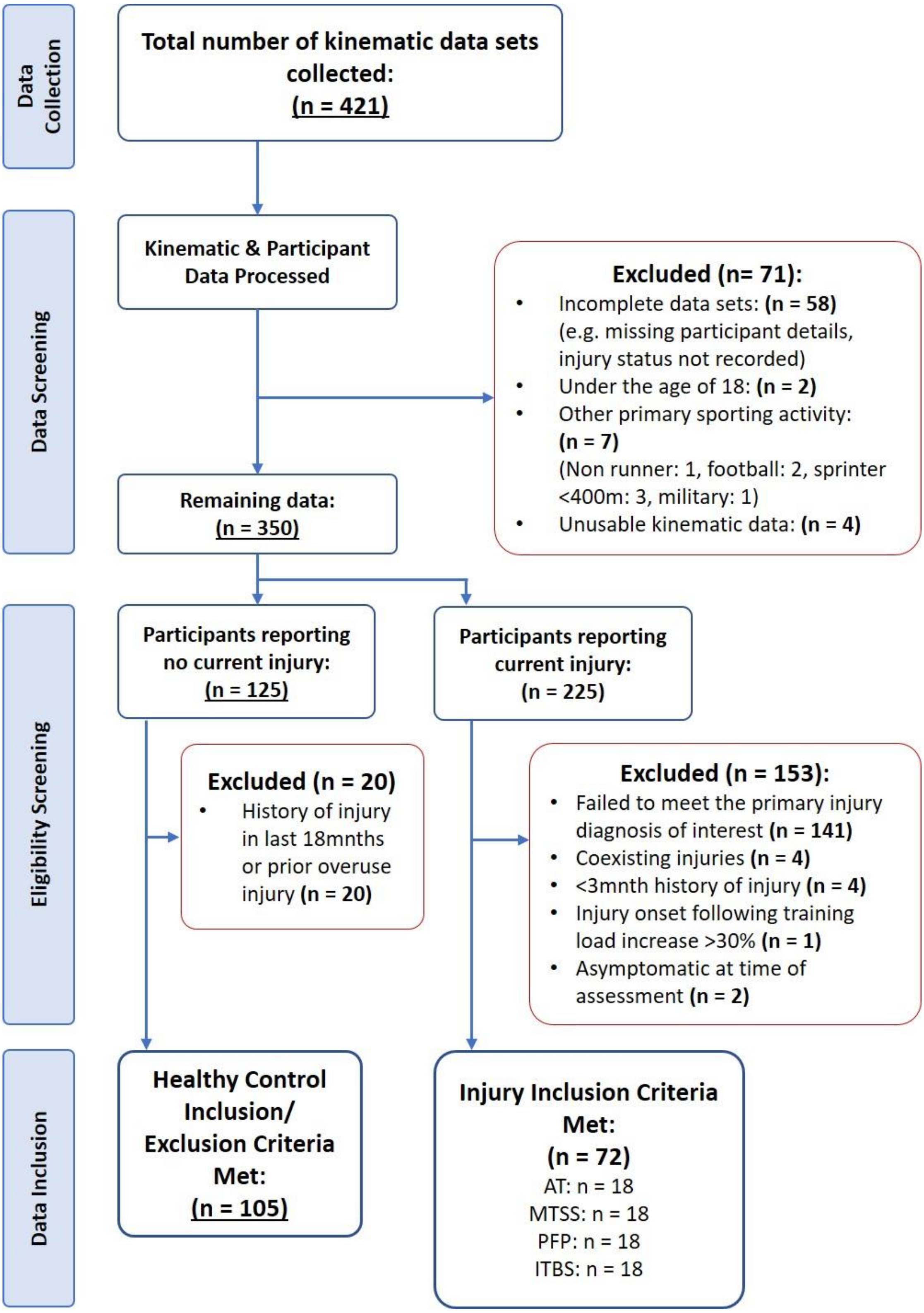
3.2.7 Total Data Collected

Between June 2013 and September 2017, kinematic data was collected from a total of 421 participants outlined in Figure 9. Seventy one datasets were excluded for reasons including: incomplete data reporting (n = 58), such as missing participant characteristics of age, mass, height, injury diagnosis or training history, errors in the kinematic data (n = 4) such as missing markers, participants under the age of 18 (n = 2) and participants for whom their primary sport was not endurance running (n = 7), including 3 sprinters competing in distance less than 400 meters, 2 footballers, 1 military recruit and 1 participant classified as a "non-runner", who did not meet the inclusion criteria of a minimum running frequency of 2 times per week. Of the remaining 350 datasets, 125 reported no current injury. Twenty of these participants were excluded due to a history of injury within the last 18 months, or having reported a history of one of the common overuse injuries of interest in this thesis (AT, MTSS, PFP or ITBS). This left a remaining 105 participants meeting the inclusion criteria for healthy control runners. Data from these participants was subsequently used to address the research questions within Chapter 4, 5 and 6.

A total of 225 participants reported a current running related injury. Following assessment by the lead clinician 153 were excluded for the following reasons: failed to meet the primary diagnosis of interest (n = 141), presented with coexisting injuries (n = 4), did not meet the minimum injury duration of greater than 3 months (n = 4), injury onset occurred following an acute increase in training volume of greater than 30% per week (n = 1) and two further participants who were asymptomatic at the time of assessment. This left a remaining 72 subjects who met the inclusion criteria for running related injuries of interest and were subsequently included within Chapter 5.

As data collection had commenced prior to the development of the specific research aims to be addressed in Chapter 7, several participants in the initial dataset who may have qualified for the gait retraining study were unable to be recruited. Once the research aims had been finalised for Chapter 7 a further ethics application was submitted and granted in July 2016 (HSCR16-49, Appendix A). Consequently, further participant recruitment took place up until 2018 in order obtain the required number of participants with PFP who met the inclusion criteria for the gait retraining study. The specific methods of this study are outlined in Chapter 7.

Figure 9: Flow chart of total numbers of kinematic datasets collected between June 2013 and September 2017.



3.3 Biomechanical procedures

In order to develop the biomechanical testing protocol, several methodological gaps were identified deemed necessary to address in order meet the aims of this thesis. Firstly, in order to track the kinematics of the trunk segment it was necessary to develop a tracking marker set. Second, to separate kinematic waveforms into multiple gait cycles it is necessary to identify gait events of foot strike and toe off. The gold standard of measurement is considered to be through force plate measurements (344). However, this thesis aimed to utilise a motorised treadmill for kinematic data collection procedures and consequently force data was not available. Therefore, there was a need to establish a method for gait event detection utilising kinematic data. Finally, during an initial search we found limited found limited evidence detailing the run duration required to achieve stable kinematic data during treadmill running. Consequently, we sought to identify an appropriate treadmill accommodation period to be utilised within the final study methodology.

In order to address these methodological gaps, several preliminary studies were conducted. These investigations were collaborative studies with the results used to inform and develop the final methodology of this thesis, as well as broader aims within the institutional research group. The findings of these studies and how they informed the final methodology are discussed in more detail within the appropriate section of this Chapter. Three of these preliminary studies have subsequently been published, with one additional, unpublished pilot study. Background publications informing the development of this methodology include:

Preece, S. J., Bramah, C., Mason, D. (2016) A marker set for measuring the kinematics of the lumbar spine and thoracic spine during running: a technical note. *Journal of Human Sport & Exercise.* 11 (3), pp. 390 – 396.

Smith, L., Preece, S., Mason, D., Bramah, C. (2015) A comparison of kinematic algorithms to estimate gait events during overground running. *Gait & Posture.* 41, pp. 39 – 43.

Mason, D. L., Preece, S. J., Bramah, C., Herrington, L. C. (2016) Reproducibility of kinematic measures of the thoracic spine, lumbar spine and pelvis during fast running. *Gait & Posture.* 43, pp. 96 – 100.

3.3.1 Kinematic data collection

Three-dimensional kinematic data was collected using a 12 camera Qualysis Oqus system sampling at 240Hz (Gothenburg, Sweden). All cameras were positioned around the laboratory in a manner that ensured each anatomical tracking maker could be visualised by a minimum of two cameras. At the beginning of each testing session a dynamic wand calibration was conducted in order to orientate the camera system within the global coordinate system and laboratory reference frame. Four reference markers attached to a ridged L- frame were positioned in the centre of the capture volume pointing in the direction of forward running progression. A 60 second calibration procedure was then conducted using a calibration wand with two retroreflective markers attached at a fixed distance apart (601.7mm). The calibration wand was systematically moved in multiple directions around the laboratory to ensure calibration of the entire capture volume of interest. Following the calibration, marker residuals of less than 0.4mm were considered acceptable, as lower residuals are associated with more accurate reconstruction of 3D marker coordinates from data collection (345).

The orientation of the laboratory coordinate system was defined using a Cartesian coordinate system following the right-hand rule in accordance with methods outlined by Grood and Suntay (346). The Z axis points vertically upward, Y axis pointing in the line of forward progression and the X axis perpendicular to the Z and Y axis. The laboratory coordinate system was subsequently used to define the segment coordinate system.

3.3.2 Marker Placement

Following calibration of the global coordinate system, a static anatomical calibration trial was conducted in accordance with the calibrated anatomical system technique (347) (248). In order to complete this trial, all static and dynamic marker were first attached to the participant and a static anatomical calibration trial was recorded with the subject stood in the centre of the laboratory facing in the direction of forward movement.

In order to track the motion of the trunk, pelvis and lower limbs retroreflective markers were attached to anatomical landmarks during static and dynamic trials. 15mm retroreflective markers were attached to the thorax, lumbar spine, pelvis, thigh, shank and foot (Figure 10, Figure 11 & Figure 12).

During preliminary testing we encountered several difficulties when establishing a marker tracking design for the trunk segment. The initial marker design trialled, consisted of a cluster of four non-colinear markers attached to the sternum, as recommended by the International Society of Biomechanics (348). However, due to the proximity of markers on the cluster, marker trajectories were found to merge within the camera field as the trunk rotated during running. This resulted in inaccurate reconstruction of marker trajectories with repeated loss of markers within the data recording. Following repeated testing, two further options were considered, both of which were clearly tracked during preliminary trials. The first method consisted of utilising three tracking markers, with two attached on bilateral acromions and one on the thorax. The second method was an adaption of the initial four marker cluster design, consisting of three non-collinear markers attached to a cluster situated on the sternum (Figure 10). These markers were attached in a way that ensured they were clearly tracked without any merging of marker trajectories.

To determine the tracking markers to be used in the final protocol, we compared the data collected using both methods which was later published as a technical note (349). The coefficient of multiple correlation was used to assess waveform similarity and the standard error of measurement averaged over the entire waveform to quantify the absolute difference between methods. The results of this investigation found poor agreement between the two marker sets in the sagittal plane with large standard error of measurements. The lack of agreement between methods was thought to be due to movement artefact of the acrominon markers, occurring due to arm movement during running (349, 350). Based on these findings, the final marker design consisted of a cluster of three non-collinear markers attached to the sternum (Figure 10). The full marker set used to track all segments followed the same protocol as outlined in Mason et al, (201) and Seay et al (330), detailed below.

The trunk segment was tracked using a rigid cluster containing 3 markers attached to the sternum (Figure 10). An anatomical reference frame for this segment was defined using markers attached to the suprasternal notch, xiphoid process, 7th cervical vertebrae and 6th thoracic vertebrae. The pelvis segment was defined using markers positioned on the iliac crest in vertical alignment with the greater trochanters. Tracking markers were placed directly over the anterior superior iliac spines and the posterior superior iliac spines and. Rigid thigh and shank clusters containing four markers were placed laterally over the thigh and shank segments and secured using double sided sticky tape and elasticated bandages. An anatomical reference frame for the thigh and shank segments was defined using calibration markers attached to bilateral greater trochanters, lateral and medial femoral condyles and lateral medial malleoli (Figure 11). To ensure clusters did not move during the testing procedures the outline of the markers were drawn onto the skin. The foot segment was defined in accordance with previous biomechanical studies modelling the foot and rearfoot (235). Tracking markers were placed directly over the shoe; 3 non-linear markers attached to the heel of the shoe to track rearfoot movement (Figure 12) and markers attached to the shoe over the base of the 5th metatarsal, 1st metatarsal and head of the 2nd metatarsal. Calibration markers for the foot and ankle segments were attached to the lateral and medial malleoli. In order to avoid the effects of intertester differences in marker application, the same examiner [CB] applied all markers (257).

Figure 10: Markers used to model the trunk segment. Static markers are labelled in yellow, tracking markers in blue.

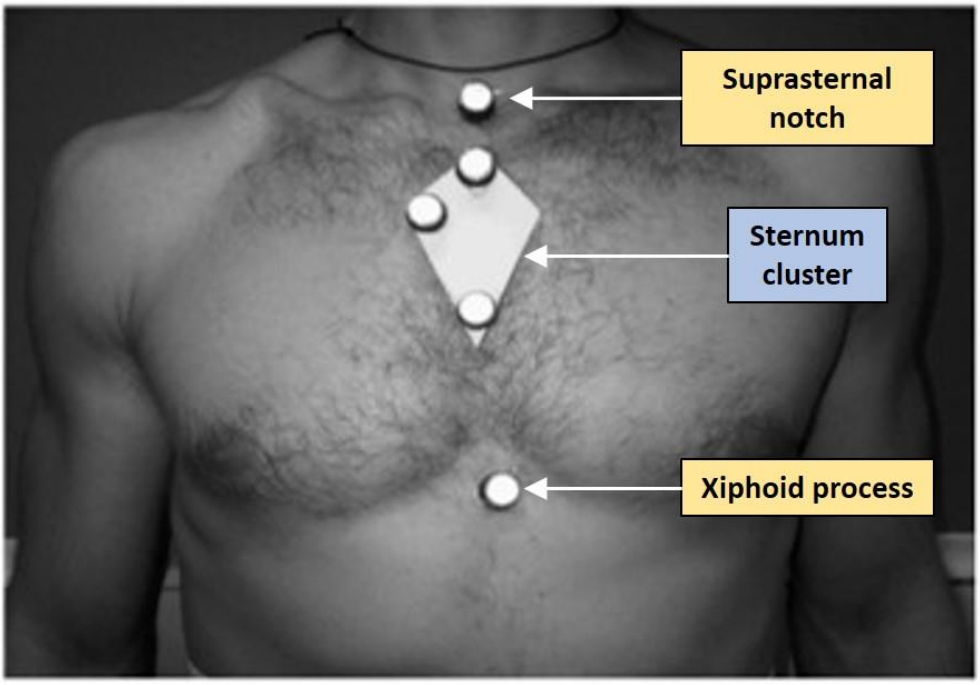


Figure 11: Markers used to model the pelvis and lower limb segments. Static markers are labelled in yellow, tracking markers in blue. PSIS = posterior superior iliac spine, ASIS = anterior superior iliac spine, GT = greater trochanter, LC = lateral condyle, MC = medial condyle. Met = metatarsal.

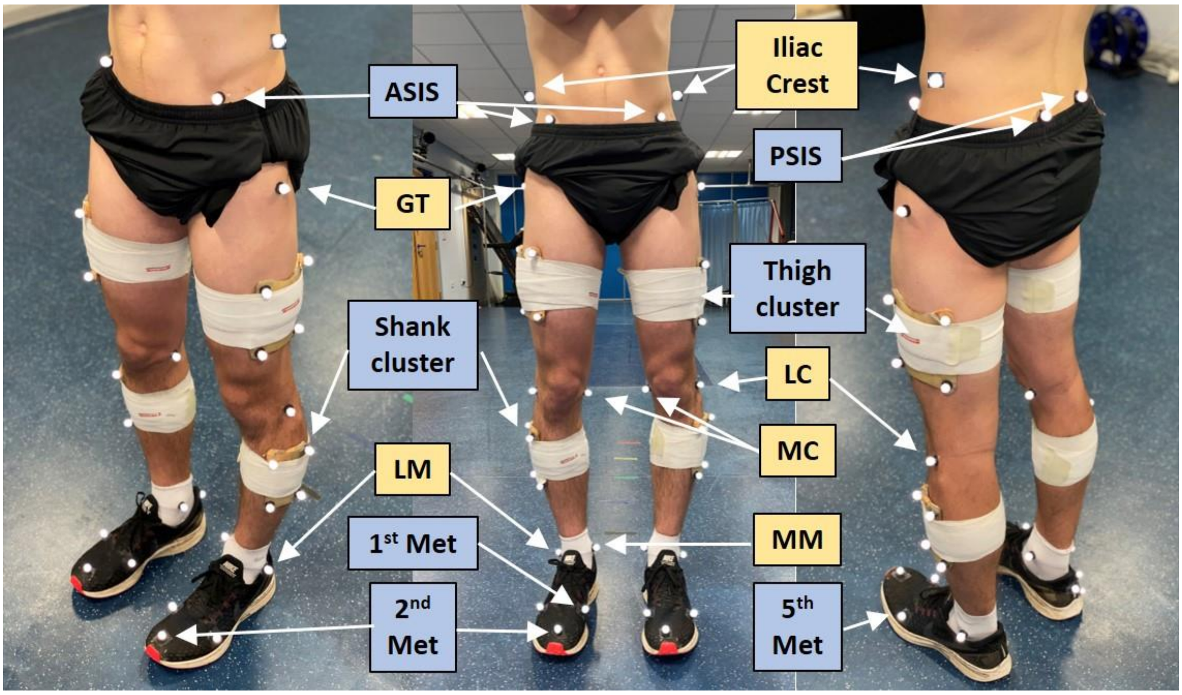
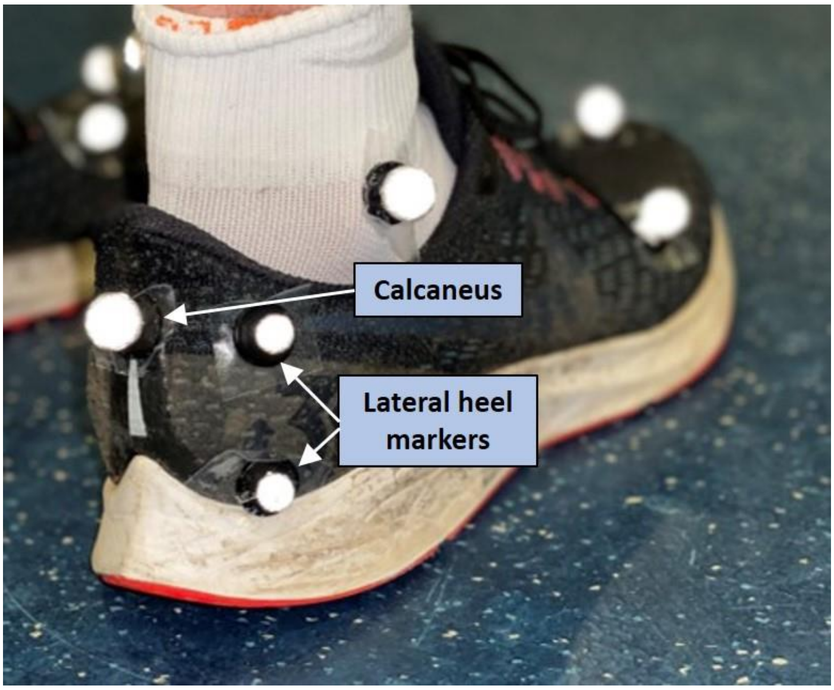


Figure 12: Markers used to track the rearfoot segment. Static markers are labelled in yellow, tracking markers in blue



3.3.3 Testing Procedures

Following the static calibration trial, calibration markers were removed prior to dynamic trials. For dynamic running trials all participants ran on a treadmill (Sole Fitness, F63, USA) at 3.2m/s using their own running shoes. Participants completed a 5-minute warm up period before 30 seconds of kinematic data was recorded in order to obtain a minimum of 10 consecutive footfalls. Where appropriate, speed was then increased to 3.8m/s, 4.5m/s and 5.2m/s for wider department research aims unrelated to this thesis. During the initial protocol development, we planned to capture running kinematics at 3.2m/s and 3.8m/s for all studies. However, during pilot testing, the large number of trials associated with multiple testing speeds resulted in increased participant perspiration and the loss of markers. Consequently, kinematic data was repeatedly unusable due to marker movement and dropouts. Therefore, the decision was made to capture runners at a single speed of 3.2m/s and restrict the gait retraining protocol to a maximum 10-minute retraining period, as opposed to 15 minutes which was initially proposed (the full gait retraining protocol can be found in Chapter 7, Section 7.2.4 &

Figure 28). The speed of 3.2m/s was selected to allow comparison to previous kinematic studies testing participants at a similar speed (58, 192).

A standardised speed was used to avoid variability in kinematic parameters that may occur due to variations in running speed (291, 351, 352). Treadmill running has been shown to reproduce highly similar kinematics throughout stance phase (353, 354) and throughout the entire gait cycle (355, 356) when compared to over ground running. Furthermore treadmill running allows subjects to achieve a constant gait pattern when running, reducing potential stride to stride variability that may occur when performing repeat running trials over an indoor running track (356). Therefore, kinematics can be considered representative of a typical over ground run. Participants wore their own running shoes in order to provide a representative measure of their normal running kinematics and avoid the effect of shoe differences on kinematics (263, 357).

During initial development of the testing procedures, we found limited found limited evidence detailing the run duration required to achieve stable kinematic data. Consequently, we conducted a preliminary study in order to determine the warm-up run duration. A total of 13 injury free participants completed continuous treadmill running for a total of 10 minutes. Thirty seconds of kinematic data were collected at 3 minutes, 5 minutes and 8 minutes during continuous running. All kinematic data was collected and processed in accordance with procedures outlined within this Section. One-way repeated measures ANOVA with a critical alpha of .05 was used to investigate differences between time points for discrete kinematic parameters at initial contact, peak angles and spatiotemporal parameters. When significant differences were observed, post hoc Bonferroni test was used to identify differences between time-points.

The results from this pilot study identified significant differences between time points for stride rate, stride length, peak ankle dorsiflexion and peak knee flexion. Pot hoc Bonferroni found differences occurred between the 3min/5min timepoint and the 3min/ 8min timepoint. No significant differences were observed between the

5min/8min timepoints. Therefore a 5-minute accommodation period was chosen for use in the final study protocol. Full results from this pilot study are provided in Appendix E.

3.3.4 Kinematic Modelling

The static calibration trial was used to define the joint coordinate system, with segments modelled as rigid bodies. The joint coordinate system for the thorax was defined similar to that outlined by the international society of biomechanics (348, 358). Specifically, this segment was defined using the z-axis (pointing upwards) as a line connecting the midpoint between markers located on the suprasternal notch and 7th cervical vertebrae, and the midpoint between the xiphoid process and 6th thoracic vertebrae. The x-axis was then defined as a perpendicular line to the plane formed between the suprasternal notch and 7th cervical vertebrae, and the midpoint between the xiphoid process and 6th thoracic vertebrae. The y-axis face anteriorly, oriented perpendicular to the x and z-axis (Figure 13). For the pelvis segment, the joint centre origin was defined using a virtual marker positioned midway between the two iliac crest markers (Figure 14A). The z-axis was aligned with the laboratory pointing upwards, the x-axis pointed from the origin to the right iliac crest marker and the y-axis pointed anteriorly, perpendicular to the z and x-axis (Figure 14A).

Figure 13: Joint coordinate system for the trunk segment. X-axis is pictured in red, y-axis in green, z-axis in blue. Static markers are highlighted in yellow, tracking markers in blue.

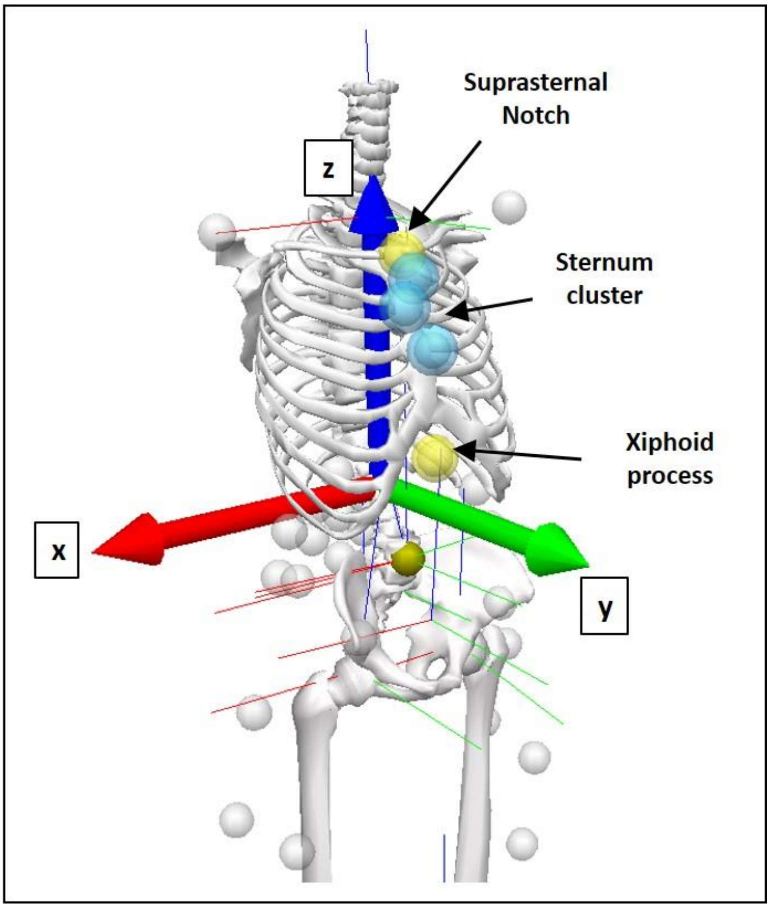
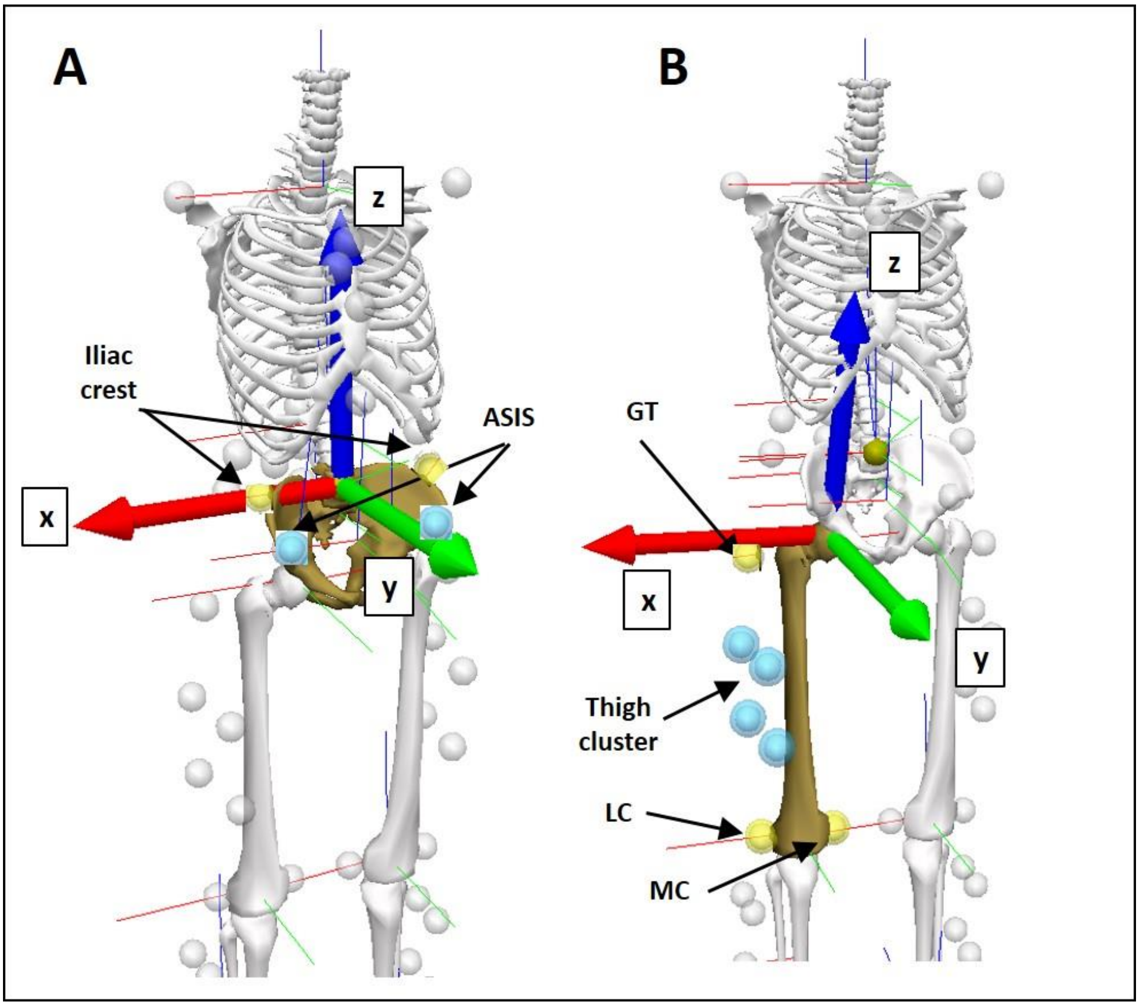


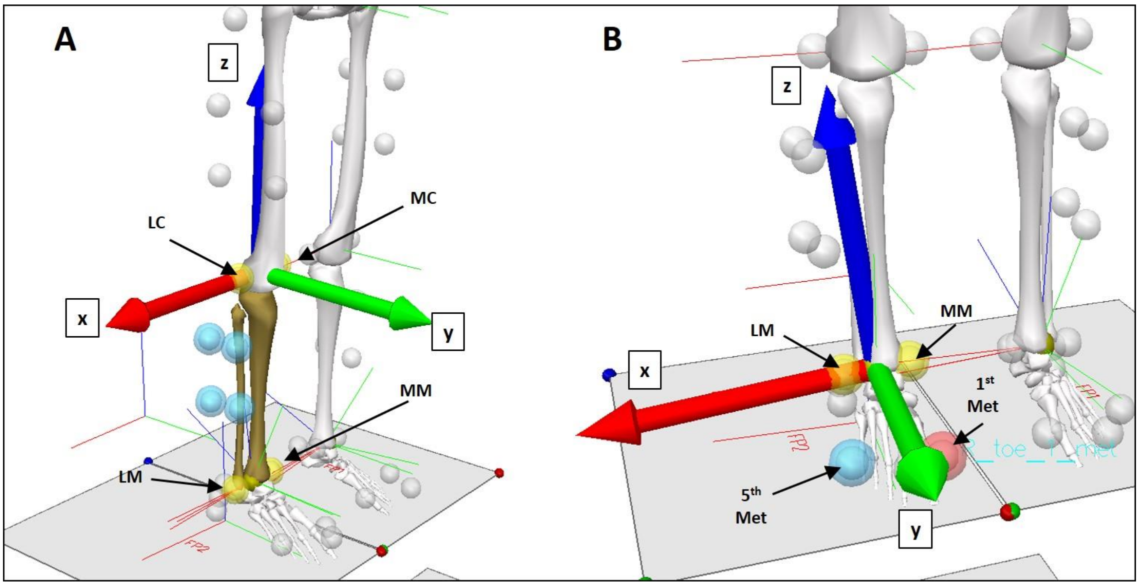
Figure 14: A: Pelvis joint coordinate system. B: Thigh joint coordinate system. X-axis is pictured in red, y-axis in green, z-axis in blue. Static markers are highlighted in yellow, tracking markers in blue. ASIS = anterior superior iliac spine, GT = greater trochanter, LC = lateral condyle, MC = medial condyle.



The hip joint centre was defined using a prediction approach based on the distance between the right and left ASIS markers and the position of the greater trochanter marker as described by Bell et al (359, 360). Specifically, the hip joint centre was located along a line projected medial to the greater trochanter marker, bisecting a point located 30% of the inter ASIS distance distal and 14% medial to the ASIS (359) (Figure 14B). The x-axis pointed towards the right, parallel to a line bisecting the ASIS markers, y-axis parallel to a line bisecting the midpoint of the PSIS and ASIS markers facing anteriorly and the z-axis pointing perpendicular to the two. The thigh was oriented with the z-axis pointing upwards aligned with the long axis of the bone, defined from the hip joint centre and the knee joint centre (Figure 14B). The x-axis pointed towards the right, orientated perpendicular to the z-axis and the y-axis facing anteriorly, perpendicular to the x and z-axis.

The shank coordinate frame was oriented with the z-axis pointing vertically, aligned with the long axis of the bone defined form the central point between the lateral and medial femoral condyles and malleoli. The x-axis of the shank was defined as the line connecting the femoral condyles, pointing towards the right and the y-axis oriented perpendicular to the z and x-axis (Figure 15). Rather than use an anatomically defined foot segment, a virtual foot segment was used in which the neutral joint ankle was defined as a flat foot with a vertical shank segment. This was to account for a slight plantarflexion offset due to the metatarsal markers being in a plantarflexed orientation relative to the malleolus markers. To define the foot segment the ankle joint origin was positioned at the midpoint of the lateral and medial malleolus markers (Figure 15B). The z-axis was oriented vertically from the ankle joint origin, the x-axis was defined as the line through the medial and lateral malleolus markers pointing towards the right and the y-axis defined as a line pointing anteriorly, through the central points between the medial/lateral malleolus and the 1st/5th metatarsal markers.

Figure 15: A: joint coordinate system for the shank. B: Joint coordinate system for the ankle. X-axis is pictured in red, y-axis in green, z-axis in blue. Static markers are highlighted in yellow, tracking markers in blue. LC = lateral condyle, MC = medial condyle, MM = medial malleolus, LM = lateral malleolus.



3.3.5 Kinematic Data Processing

Marker trajectories were first labelled in Qualysis Track Manager (Gothenburg, Sweden), before being exported into Visual 3D where the raw kinematic data were low pass filtered at 10Hz. A 10Hz filtering frequency has been reported to effectively filter signal noise while still providing stable kinematic waveforms (361).

During the initial development of the data processing methods we initially trialled a global optimisation modelling technique for reconstruction of segmental kinematics. However, during data processing attempts, the global optimisation model significantly increased the time required for data processing, beyond that which was considered acceptable. Global optimisation has previously been proposed as a modelling technique to minimise the effects of soft tissue artefact on segmental kinematics by imposing specific joint constraints, limiting translational movements between segments (362). Whereas alternative models, such as the 6 degrees of freedom (6DOF) model, do not impose joint constraints and are thought to be more susceptible to kinematic errors due to soft tissue artefact (362). However, in a previous study, we compared the reproducibility and standard error of measurement between global optimisation and 6DOF models, finding similar levels of reproducibility between methods (201). Therefore, a 6DOF model was considered acceptable for calculation of segmental kinematics.

Intersegmental kinematics, along with the motions of the pelvis and thorax with respect to the laboratory system, were subsequently calculated in visual 3D using a 6DOF model. A cardan angle sequence was used to define joint orientation using a right-hand rule and joint angle conversion of x-y-z. With x = flexion/extension, y = abduction/ adduction, z = internal/ external rotation. The zero position for all joint angles were derived from the static trial.

3.3.6 Gait event detection

In order to separate kinematic waveforms into multiple gait cycles it is necessary to identify gait events of foot strike and toe off. The gold standard of measurement is considered to be through force plate measurements (344). However, this thesis aimed to utilise a motorised treadmill for kinematic data collection procedures and consequently force data was not available. Therefore, there was a need to establish a method for gait event detection.

In order to identify initial contact and toe off we conducted a prior investigation to develop a kinematic algorithm for gait event detection during running (344). The initial study compared gait events derived from kinematic data to those derived using ground reaction force data during over ground running. The proposed algorithm estimated initial contact as the first maxima between heel and PSIS markers, while toe off was estimated using the maximal displacement between 2nd metatarsal marker and PSIS. Results identified this algorithm to display a root mean square (RMS) error of 14.1ms and 9.2ms for initial contact and toe off respectively (344). However, a subsequent kinematic algorithm was later published by Handsaker et al (363) reporting lower root mean square error (RMS) values of 8.3ms and 5.6ms for initial contact and toe off respectively. This kinematic algorithm has subsequently been used by additional studies investigating treadmill running kinematics (364) and was therefore chosen for use within this thesis.

The final algorithm utilised to determined gait events of initial contact and toe off was defined using the kinematic approach reported by Handsaker et al (363). Using this approach initial contact was defined as the first vertical acceleration peak of either the heel or metatarsal markers and toe off defined as the vertical jerk peak of the 2nd metatarsal marker (363).

3.3.7 Derivation of Kinematic Parameters

Kinematic curves collected for a continuous period during running were segmented into separate gait cycles using the gait event detection algorithm detailed above. Gait events were subsequently used to segment each kinematic signal into a minimum of 10 consecutive gait cycles. A minimum of 10 consecutive gait cycles were recorded as greater reliability of kinematic measurements has been reported when data are averaged over several running trials (254). An ensemble average for each signal was created and selected kinematic parameters derived from the ensemble average curves. This latter processing was carried out using a custom Matlab script.

A range of kinematic parameters were analysed including joint angles at initial contact, peak stance phase angles and joint excursions of the trunk, pelvis, hip, knee and ankle. Parameters were selected for analysis based on previous literature identifying kinematic characteristics associated with running related injuries as well as gaps identified within the current literature (Section 2.1.7.2). Peak angles at during stance were defined as the maximum joint angle between initial contact and toe off and joint excursions were defined as the total range of movement from initial contact to the peak angle.

4 Chapter 4 – The between day repeatability, standard error of measurement and minimal detectable change for discrete kinematic parameters during running.

The aim of this Chapter was to establish the repeatability of discrete kinematic parameters during running. Within Chapter 2 of this thesis, the literature review identified several discrete kinematic parameters to be associated with common running injuries, which are subsequently the focus of intervention studies utilising test-retest designs. By establishing the repeatability of discrete kinematic parameters, the results from this Chapter were deemed necessary to aid interpretation of between-group and between day differences in running kinematics. The presented results were subsequently used to evaluate the robustness of the kinematic data collected as part of subsequent chapters of the thesis (Chapter 5 & 7).

4.1 Introduction

Running kinematics have been proposed as an intrinsic risk factor for running related injuries. In Section 2.1.7.2 several kinematic parameters during the stance phase of running were found to be associated with common running related injuries. These parameters include frontal plane pelvis kinematics, as well as frontal and transverse plane kinematics of the hip, knee and ankle. It is thought that these kinematic parameters increase the stress placed on the musculoskeletal system during each foot contact of a run, leading to cumulative tissue overload and injury development. Consequently, several studies have sought to investigate whether clinical interventions can improve running kinematics and clinical outcomes amongst injured runners.

Gait retraining has been proposed as a clinical intervention which aims to correct abnormal running kinematics (49, 66). Using test-retest designs, several studies have investigated the effects of gait retraining on kinematics amongst injured runners. Many of these studies have reported post intervention reductions in frontal and transverse plane kinematics at the hip and pelvis (69, 70, 255). However, only one study has provided data regarding the test-retest reliability and the standard error of measurement associated with their testing procedures (70).

Current literature suggests between day repeatability of running kinematics is generally poor, yielding potentially large measurement errors. This is particularly true for frontal and transverse plane kinematics of the pelvis, hip and lower limbs (199, 201, 253). Peak frontal plane pelvis kinematics have been reported to demonstrate measurement errors of up to 1.7^{o} (201), whereas peak hip adduction and internal rotation angles have been shown to demonstrate measurement errors ranging from 0.97^{o} (252) to 2.7^{o} (199) and 1.1^{o} (252) to 5.9^{o} (253) respectively. This has implications for many intervention studies reporting changes in hip and pelvis kinematics. Specifically, without adequate reporting of the measurement error associated with testing procedures, it is difficult to ascertain whether kinematic changes are the result of intervention effects or between day measurement error. Consequently, this could result in the interpretation of results as being "meaningful" when they are instead the result of error in measurements (256).

It is important to note, that the repeatability of testing procedures has also been shown to vary across laboratories. This is highlighted by results from Noehren et al (252) and Stoneham et al (253), reporting SEM values for peak hip internal rotation of 1.1^{o} and 5.9^{o} respectively. Reasons for the differences in reported measurement errors could be explained by between laboratory differences in kinematic testing procedures. For example, Noehren et al (252) tested participants during treadmill running whereas Stoneham et al, (253) utilised over ground running test procedures. Furthermore, marker reapplication errors have been shown to produce large errors in kinematic data (261) which can vary between and within examiners (257). Therefore, reporting the repeatability for individual laboratory testing procedures is recommended to aid interpretation of findings and account for between laboratory differences in testing procedures.

Currently no study has reported the repeatability of trunk and pelvis kinematics during treadmill running. Many gait retraining studies utilise treadmill testing procedures to investigate between day differences in discrete kinematic parameters of the trunk (267), pelvis and lower limbs (255). However, as there is no prior data reporting the measurement error associated with trunk and pelvis kinematics during treadmill running, it is difficult to identify whether post intervention differences represent true intervention effects or are the result of measurement error.

4.1.1 Aim and Objective

The overall aim of this Chapter was to establish the repeatability of discrete kinematic parameters during running in order to aid the interpretation of between-group and post intervention kinematic differences within subsequent chapters of this thesis. In order to achieve this aim, the specific objective of this study was to investigate the between day repeatability, standard error of measurement and minimal detectable change of discrete kinematic parameters of the trunk, pelvis and lower limbs during treadmill running.

4.2 Methods

4.2.1 Participants

A total of 16 injury free control participants were included within this study (Table 21). Participants were recruited via poster advertisements at local running clubs and sports injury clinics. Participants were included providing reported no injury within the last 18 months. Participants were excluded if they reported any musculoskeletal ailment within the last 18 months that caused a restriction or cessation of running, or any need to seek advice from a health care professional. Exclusion criteria included any current or previous history of overuse running injury, injury caused by another sport, any previous spinal injury or lower limb surgery (Section 3.2.1.2). Additional exclusion criteria included any newly sustained injury or attempt to change their gait between tests 1 and 2. All participants provided written informed consent prior to participation and ethical approval was obtained via the local ethics committee.

Table 21: Participant characteristics. Mean [SD].

Male/ Female	Age (years)	Mass (kg)	Height (cm)	BMI (kg/m^2)	Run Frequency (runs per week)	Average Weekly Run Volume (Miles)
6/ 10	34.4 (10.2)	59.7 (10.8)	169.1 (9.4)	20.7 (2.2)	6.5 (2.9)	44.5 (26.6)

4.2.2 Kinematic data collection

All participants were required to attend two data collection sessions two weeks apart. At each testing session kinematic data were collected from all participants whilst running on a treadmill at 3.2m/s in accordance with methods outlined in Section 3.3. Participants were instructed to continue their normal training routines between testing sessions. At each session a 5-minute warm up period was provided, after which, 30 seconds of kinematic data were collected using a 12 camera Qualysis Oqus system (240Hz). Anatomical segments of the trunk, pelvis, bilateral thighs, shank and feet were tracked using retroreflective markers attached to anatomical landmarks. Full details of the markers used to track each segment and the precise definition of the anatomical coordinate systems is provided in Section 3.3.

Raw kinematic data were low pass filtered at 10Hz. Intersegmental kinematics, along with the motions of the pelvis and thorax with respect to the laboratory system, were calculated using a 6DOF model in Visual 3D (C-Motion). Gait events were defined using a kinematic approach (363) and subsequently used to segment each kinematic signal into a minimum of 10 consecutive gait cycles. An ensemble average for each signal was created and selected kinematic parameters derived from the ensemble average curves. This latter processing was carried out using a custom Matlab script.

4.2.3 Data Analysis

Several discrete kinematic parameters commonly reported in kinematic investigations were selected for analysis. These included sagittal, frontal and transverse plane kinematics of the trunk, pelvis, hips, knees and ankles at initial contact, peak angle and stance phase joint excursions. Parameters at initial contact included sagittal plane angles of the trunk, pelvis, hip, knee and ankle as well as frontal plane angles of the trunk and rearfoot. Peak angles at mid stance included sagittal and frontal plane angles of the trunk, pelvis, knee and ankle and rearfoot as well as transverse plane angles of the hip and knee. Parameters were selected based on those identified within the literature review as being associated with common running injuries, along with parameters with limited prior research (Section 2.1.7.2, Table 18). Peak angles were defined as the maximum joint angle between initial contact and toe off.

4.2.4 Statistical Analysis

In order to assess the between day repeatability of kinematic parameters, the interclass correlation coefficient (365) was first calculated. The ICC was chosen as the statistical method of use as this method reflects both the degree of correlation and consistency between results (366). This is in contrast to alternative methods such as Pearson's correlation coefficient, which quantifies the degree of correlation between two measurements, or Bland-Altman plots which reflects only the level of agreement (265, 366, 367). The use of ICC was selected to permit comparison of results between the present study and previous biomechanical studies (199, 252). However, ICC values alone are of limited clinical value, as they do not provide estimates of the measurement precision in units specific to the measurement system. Therefore, the interclass correlation coefficient (365) was calculated along with the standard error of measurement (SEM) and the minimal detectable change (MDC).

4.2.4.1 Interclass Correlation Coefficient (365)

ICC estimates and their 95% confidence intervals were calculated using SPSS (IBM Statistics v23) (SPSS Inc, Chicago, IL) using a two-way mixed effects model, mean of k measurements with absolute agreement (366). ICC with absolute agreement was selected over the ICC method with consistency, as ICC with consistency does not consider systematic differences between measurements and therefore may lead to an overestimation of the reliability of measurement (367).

Interclass correlation coefficient provides a value ranging between 0 (equalling no reliability) to 1 (equalling perfect reliability) indicating the level of agreement between two measurements (264, 265). Values of <0.5, 0.5 to 0.75, 0.75 to 0.9 and >0.9 were interpreted as poor, moderate, good and excellent respectively (366).

4.2.4.2 Standard Error of Measurement (SEM)

The SEM is considered an estimate of the expected variation in scores that may occur due to random error and therefore can be used to provide an estimate as to the precision of measurement reported in the unit of the measurement (264, 266). The SEM represents the 68% confidence interval for a set of scores. The standard error of measurement was calculated as:

$$SEM = SD \times \sqrt{(1 - ICC)}$$

4.2.4.3 Minimal Detectable Change (MDC)

The minimal detectable change provides the minimal threshold beyond the random measurement error with a 95% confidence interval. Therefore, minimal detectable change is considered to represent the degree of change representative of a true change, greater than that which could be explained by random error (266). MDC is calculated from the SEM and a degree of confidence using the multiplier of square root of 2. This is to account for any additional uncertainty introduced by using different scores from measurements of two time points (266, 368). Minimal detectable change values were calculated as:

$$MDC = SEM \times 1.96 \times \sqrt{2}$$

4.3 Results

Kinematic data at initial contact, peak angles and excursions along with the ICC values, 95% confidence intervals, SEM and MDC are presented in Table 22, Table 23 and Table 24.

4.3.1 Initial contact

At initial contact frontal and sagittal plane kinematic parameters of the trunk, pelvis, hip, knee and ankle were found to demonstrate ICC values ranging from 0.839 to 0.941

representing good to excellent repeatability (Table 22). SEM values were relatively low, ranging from 0.6⁰ for frontal plane pelvis angle and 2.6⁰ for frontal plane rearfoot angle at initial contact. ICC values ranged from 0.525 to 0.77 for transverse plane kinematics of the hip and knee representing moderate repeatability.

Table 22: Between day repeatability of kinematic parameters at initial contact. Interclass correlation coefficient [ICC] values of <0.5, 0.5 to 0.75, 0.75 to 0.9 and >0.9 were interpreted as poor, moderate, good and excellent respectively. SEM = Standard Error of Measurement, MDC = Minimal Detectable Change. Mean [SD] values represent degrees.

Initial Contact								
Parameter		Mean (SD)		ICC	95%CI		SEM (°)	MDC (°)
		Day 1	Day 2		lower	upper		
Trunk	Forward Lean	5.3 (5.7)	5.6 (4.4)	0.866	0.614	0.953	1.8	5.1
	Ipsilateral Flexion	2.9 (2.8)	2.9 (2.6)	0.910	0.739	0.969	0.8	2.2
Pelvis	Anterior Tilt	8.1 (5.2)	8.0 (5.7)	0.928	0.791	0.975	1.5	4.0
	Contralateral Pelvic Drop	2.2 (1.4)	2.2 (1.3)	0.829	0.499	0.941	0.6	1.5
Hip	Flexion	23.5 (5.1)	23.9 (6.0)	0.941	0.832	0.979	1.3	3.7
	Adduction	5.6 (2.9)	5.9 (3.1)	0.883	0.668	0.959	1.0	2.8
	Internal Rotation	0.3 (4.3)	2.0 (6.0)	0.633	0.003	0.870	3.2	8.8
Knee	Flexion	5.5 (5.8)	7.2 (6.6)	0.839	0.555	0.943	2.5	6.9
	Adduction	0.9 (3.1)	0.9 (3.5)	0.925	0.785	0.971	0.9	2.5
	Internal rotation	5.3 (6.0)	4.7 (6.4)	0.770	0.331	0.920	2.9	8.1
Foot/ Ankle	Ankle Dorsiflexion	5.4 (8.4)	4.0 (8.9)	0.914	0.761	0.970	2.5	6.9
	Rearfoot Inversion	7.0 (6.3)	8.0 (8.2)	0.868	0.628	0.954	2.6	7.2

4.3.2 Peak Angles

Several peak angles at mid stance demonstrated excellent between day repeatability with low SEM values (Table 23). Specifically, peak trunk ipsilateral flexion, anterior pelvic tilt, contralateral pelvic drop, hip adduction, and ankle dorsiflexion all demonstrated excellent repeatability with ICC values greater than 0.9 and SEMs ranging from 0.6^{o} to 1.1^{o}. Transverse plane kinematics of the hip and knee demonstrated moderate to good repeatability with ICC values of 0.739 for peak knee external rotation and 0.783 for peak hip internal rotation. Although peak hip internal rotation angle demonstrated good between day repeatability, the largest SEM and MDC were observed for this parameter with an SEM of 3.2^{o} and MDC of 8.7^{o}.

Table 23: Between day repeatability of peak kinematic parameters. Interclass correlation coefficient [ICC] values of <0.5, 0.5 to 0.75, 0.75 to 0.9 and >0.9 were interpreted as poor, moderate, good and excellent respectively. SEM = Standard Error of Measurement, MDC = Minimal Detectable Change. Mean [SD] values represent degrees.

Peak Angles								
Parameter		Mean (SD)		ICC	95%CI		SEM (°)	MDC (°)
		Day 1	Day 2		lower	upper		
Trunk	Forward Lean	11.2 (6.3)	10.1 (5.2)	0.799	0.438	0.929	2.6	7.1
	Ipsilateral Flexion	4.5 (2.3)	4.1 (2.6)	0.914	0.761	0.97	0.7	2.0
Pelvis	Anterior Tilt	6.2 (4.7)	6.1 (5.1)	0.946	0.846	0.981	1.1	3.1
	Contralateral Pelvic Drop	4.7 (2.4)	4.9 (1.8)	0.917	0.767	0.971	0.6	1.7
Hip	Adduction	11.6 (2.8)	11.9 (2.7)	0.941	0.836	0.979	0.7	1.8
	Internal Rotation	3.3 (5.9)	5.8 (7.5)	0.783	0.399	0.923	3.2	8.7
Knee	Flexion	31.4 (3.5)	31.5 (5.1)	0.825	0.489	0.939	1.8	5.0
	Abduction	1.3 (2.9)	1.9 (3.4)	0.826	0.516	0.938	1.3	3.6
	External rotation	6.2 (5.0)	7.9 (6.8)	0.739	0.281	0.907	3.0	8.4
Foot/ Ankle	Ankle Dorsiflexion	21.1 (3.0)	21.3 (3.0)	0.938	0.825	0.978	0.7	2.0
	Rearfoot Eversion	2.7 (4.4)	3.2 (5.9)	0.804	0.433	0.932	2.3	6.3

4.3.3 Joint Excursions

ICC values for joint excursions demonstrate good to excellent repeatability for all parameters with a ICCs ranging from 0.774 for ankle dorsiflexion to 0.984 for trunk forward lean (Table 24). Frontal plane pelvis excursion demonstrated the lowest SEM and MDC with knee flexion excursion demonstrating the highest. SEM's and MDC's ranged from 0.5^{o} to 2.2^{o} and 1.3^{o} to 6.0^{o}.

Table 24: Between day repeatability of joint excursion. Interclass correlation coefficient [ICC] values of <0.5, 0.5 to 0.75, 0.75 to 0.9 and >0.9 were interpreted as poor, moderate, good and excellent respectively. SEM = Standard Error of Measurement, MDC = Minimal Detectable Change. Mean [SD] values represent degrees.

Joint Excursion								
Parameter		Mean (SD)		ICC	95%CI		SEM (^{o})	MDC (^{o})
		Day 1	Day 2		lower	upper		
Trunk	Forward Lean	9.8 (5.7)	10.3 (5.5)	0.984	0.954	0.994	0.7	1.9
Trunk	Ipsilateral Flexion	2.1 (1.5)	2.3 (1.8)	0.827	0.513	0.939	0.7	1.9
Pelvis	Anterior Tilt	4.5 (2.3)	4.4 (2.5)	0.883	0.66	0.959	0.8	2.3
Pelvis	Contralateral Pelvic Drop	3.0 (2.0)	3.2 (2.2)	0.951	0.864	0.983	0.5	1.3
Hip	Adduction	6.2 (2.2)	6.3 (2.3)	0.826	0.49	0.939	0.9	2.6
Hip	Internal Rotation	3.7 (3.4)	4.4 (3.7)	0.852	0.59	0.948	1.4	3.8
Knee	Flexion	26.0 (5.1)	24.4 (6.1)	0.850	0.582	0.947	2.2	6.0
Knee	External Rotation	11.8 (4.4)	12.9 (3.3)	0.782	0.403	0.923	1.8	5.0
Foot/ Ankle	Ankle Dorsiflexion	19.3 (4.2)	20.2 (4.9)	0.774	0.366	0.921	2.1	5.9
Foot/ Ankle	Rearfoot Eversion	9.8 (3.1)	11.2 (3.4)	0.885	0.529	0.964	1.1	3.1

4.4 Discussion

The objective of this Chapter was to investigate the between day repeatability, standard error of measurement and minimal detectable change of discrete kinematic parameters of the trunk, pelvis and lower limbs during treadmill running. Good to excellent repeatability was observed for sagittal and frontal plane kinematics at initial contact, peak angles during stance and joint excursions, while transverse plane kinematics tended to demonstrate lower between day reliability with large SEM and MDC values.

The findings of the present study are in agreement with those of several previous studies, in that sagittal and frontal plane kinematics tend to be more repeatable than those in the transverse plane (199, 201, 253). In particular, peak transverse plane hip and knee kinematics were observed to demonstrate the lowest ICCs and highest SEMs of all parameters (Table 23). Transverse plane kinematics are considered to be the most vulnerable to measurement errors, which perhaps explains the low repeatability when compared to sagittal and frontal planes. Therefore, the interpretation of transverse plane kinematics should be done so with caution, as large measurement errors suggest a high level of noise present within this data. The consequence of this is that it may lead to inaccurate conclusions regarding between day differences in running kinematics, as well as induce large variability within group level data. This may subsequently reduce statistical power to detect small, potentially meaningful between-group differences.

One source of error may occur through between day errors in marker reapplication. Marker reapplication inaccuracies are considered to produce the largest source of error in kinematic measurements (249, 258). Subtle misplacements in static anatomical reference markers can offset joint centre locations, resulting in altered segment orientations upon 3D reconstruction (258). Consequently, angular joint rotations can be dramatically over or underestimated, with transverse plane kinematics reported to be more vulnerable than frontal or sagittal plane (370). Osis et al (370) reported that as little as a 10mm offset of the lateral knee joint marker in the anterior–posterior direction produced errors of up to 4.8^{o} and 5.1^{o} in peak transverse plane hip and knee angles, whereas sagittal plane errors were considered relatively low, of only 1.6^o for peak knee flexion and 0.8^o for peak ankle dorsiflexion (370).

In the current study we attempted to control for marker placement errors by ensuring the same experience examiner positioned all the static markers (257). Despite this, a degree of error is still clearly evident within the current testing procedures highlighted by the observed SEMs for transverse plane kinematics (Table 22, Table 23 & Table 24). That said, only one previous study has reported a lower SEM for transverse plane hip kinematics during running, reporting an SEM of 1.1^o (252). However, this was following the use of a marker reapplication devise, designed to measure and record the precise location of anatomical reference markers which was not available in the present study. Nonetheless, this highlights the need for methods of improving the accuracy of marker placement in order to produce greater between day repeatability of kinematic measurements.

Despite lower repeatability of transverse plane kinematics compared to that of other planes, the observed repeatability values appear greater than several previous studies. Specifically, peak hip internal rotation and hip adduction were observed to demonstrate good and excellent repeatability, with ICCs of 0.78 and 0.94 respectively. Conversely, previous studies have reported ICCs of only 0.54 (199) and 0.6 (253) for peak hip internal rotation and 0.69 for peak hip adduction (199, 253). Similar observations were made for several other parameters including peak rearfoot eversion and knee abduction, demonstrating good repeatability compared to only moderate reliability values reported elsewhere (199, 253).

One explanation for the greater repeatability observed in the present study could be due to the use of treadmill testing procedures. Many prior studies have investigated repeatability of kinematics during over ground running which could induce greater movement variability between trials (199, 201, 253). This may occur due to subtle variations in running speed, air resistance or targeting of force plates during over ground running (356). In a previous study following the same kinematic testing procedures, we reported the between day repeatability during over ground running, with SEMs of 1.7^o

and 2.3^{o} observed for frontal plane pelvis and hip kinematics (201). These values are greater than those reported in the present study of 0.6^{o} and 0.7^{o}, suggesting that repeatability of running kinematics may be improved during treadmill running. However there were differences in the statistical methods between studies as measurement error was calculated as an average across the entire kinematic waveform in the prior study (201), compared to that of discrete parameters in the present study.

Only one previous study has reported the repeatability of kinematic testing procedures during treadmill running (252). In a group of 10 healthy subjects Noehren et al (252) investigated the between day repeatability of kinematics during treadmill running. Although they did not report pelvis and trunk kinematics, they did report discrete lower limb kinematics with SEMs of 3.8^{o} and 0.9^{o} for peak hip internal rotation and adduction respectively. These values are similar to the SEMs of 3.2^{o} and 0.7^{o} reported in the current study and lower than that of several previous over-ground investigations (199, 201, 253). Collectively, these results suggest that the between day repeatability of kinematics may be improved during treadmill running. However, future studies should consider directly comparing the two.

In contrast to previous studies, we reported the minimal detectable change (MDC) for a range of kinematic parameters. In the data presented, large MDC values were observed for several of the studied parameters. These included transverse plane knee kinematics at initial contact, peak rearfoot eversion, trunk forward lean, knee external rotation and hip internal rotation. Interestingly, many of these parameters are frequently implicated in the aetiology of running related injuries and subsequently the focus of clinical interventions. However, observed intervention effects can often be relatively small, with reductions in peak hip adduction angles reported to range between 1.7^{o} and 2.4^{o} following step rate manipulation (71, 255), and reductions in peak hip internal rotation of up to 5.1^{o} (255). In many instances, the magnitude of kinematic differences observed would fail to exceed the MDC values of 8.7^{o} and 1.8^{o} for peak hip internal rotation and hip adduction reported in the present study.

The consequence of these errors is that it may lead to the misinterpretation of subtle between day differences in running kinematics as clinically meaningful when they are in fact due to measurement error. Inherent to any measurement system is a degree of error, which needs to be accounted for if appropriate interpretations of between day differences are to be made. These errors are likely to vary between laboratories based on the testing procedures utilised and populations studied and may not be accurately represented by commonly used repeatability measures such as ICCs (249). Furthermore, although the statistical use of the ICC allows interpretation of the repeatability of measurements, values presented do not provide estimates of the measurement precision and therefore have limited clinical utility (249). In contrast, the SEM provides an estimate of the expected variation in scores that may occur due to measurement error, while the MDC provides the minimal threshold beyond which measurement error is expected to occur. Therefore the MDC could be considered the degree of change required to represent a true difference (266). As such, the MDC's presented in the present study are to be used as a reference point to assist the interpretation of between-group and post intervention kinematic differences reported within subsequent chapters of this thesis.

4.4.1 Limitations

There are several limitations to this study that should be acknowledged. Firstly, using a convenience sample it is possible that participants included may not be representative of wider running populations. In attempt to limit sampling bias induced through convenience sampling, we recruited participants from a range of locations frequented by recreational runners, including local running clubs, running race events and sports injury clinics (Section 3.2). This methodology was also employed throughout subsequent chapters of the thesis. Consequently, we are confident that the data on repeatability presented in this chapter are appropriate for the interpretation of data presented throughout this thesis.

While we are confident that the repeatability data form an appropriate benchmark for this thesis, we do acknowledge that the specific sample studied may possess characteristics which subsequently limit generalisability to wider running populations.

One such example is the body mass of participants included within the present study. As the average participant body mass index of 20.7kg/m^2 could be considered relatively low. Consequently, it is likely that these participants had significantly less body fat compared to other recreational running populations, such as novice runners. Considering soft tissue artefact is reported to produce large measurement errors for transverse plane kinematics, the level of error in the present study may be much lower than that in previous or future studies utilising participants with a greater body mass index.

Second, all participants were considered experienced runners, having met the inclusion criteria of a minimum two years running experience. It is possible that experienced runners may demonstrate more stable running patterns, with less movement variability, acquired through regular endurance running (371). This is in contrast to novice or injured runners, who may be more variable in their movement patterns. Nevertheless, previous studies have either reported injured runners to demonstrate less movement variability when compared to injury free populations (372), or failed to identify any difference in kinematic variability between injured runners and controls (373). Therefore, we feel it is unlikely that differences in movement variability would influence the results of the present study. However, future studies should consider investigating the repeatability of discrete kinematic parameters amongst injured populations.

Finally, due to the lack of assessor blinding there is the possibility for recall bias regarding maker placements. However, steps were taken to mitigate this effect by conducting tests a minimum of two weeks apart. This ensured that any residual traces of marker placement would be highly unlikely to be present at follow up testing.

4.5 Summary and Implications

The results from this study highlight the between day repeatability as well as the standard error of measurement and minimal detectable change of discrete kinematic parameters during the stance phase of running. This is the first study detailing the measurement error and MDC for discrete kinematic parameters of the trunk and pelvis during treadmill running. Considering stance phase kinematics are associated with common running injuries and the target of clinical interventions, the reported values will be used throughout subsequent chapters in order to identify whether kinematic differences are representative of true between-group differences and intervention effects.

5 Chapter 5 – Is there a pathological running gait associated with common running injuries?

The aim of this Chapter was to conduct a case-control study to investigate whether similar kinematic parameters are associated with multiple different running related injuries. The rationale for this Chapter was based on the findings of the literature review presented in Chapter 2. Specifically, Chapter 2 identified a number of kinematic parameters which have been associated with multiple different running related injuries. Suggesting these parameters may increase tissue loading throughout the musculoskeletal system and could represent global kinematic characteristics associated with running injuries.

The results of this Chapter were subsequently used to focus chapters 6 and 7, exploring whether the kinematic parameters are associated with weekly training load exposure (Chapter 6) and whether gait retraining interventions targeted to these kinematics, can improve kinematics, clinical and functional outcomes amongst injured runners.

Following peer review, the results of this Chapter have been published within the American Journal of Sports Medicine (Appendix F). The following account includes an extended discussion of the published work:

Bramah, C., Preece, S, J., Gill, N., Herrington, L. (2018) Is there a pathological running gait associated with common soft tissue running injuries? American journal of Sports Medicine, 46 (12), pp 3023 – 3031.

5.1 Introduction

Running is an increasingly popular method of physical activity, however it also poses a risk of injury to the musculoskeletal system. It has been reported that approximately 50% of runners become injured annually with 25% injured at any one time (11). The majority of running related injuries are considered to be overuse injuries, with the most frequently injured sites including the knee, foot and lower leg, with incidence rates reported of around 50%, 39% and 32% respectively (7). Less common injury sites include the ankle and lower back, as well as the hip and pelvis, with incidence rates ranging from 4% to 16%, 5% to 19% and 3 to 11% respectively (374). Of all running related injuries, the most frequently cited injuries include patellofemoral pain, iliotibial band syndrome, medial tibial stress syndrome, Achilles tendinopathy, plantar fasciitis, stress fractures and muscle strains (12, 13). Many of these injuries are known to have high reoccurrence rates, leading to a reduction or cessation of training in approximately 30 to 90% of cases (17). The factors related to the development of running related injuries are multifactorial and diverse, however it is widely accepted that abnormal running kinematics play a role (53, 59, 67).

There has been a large amount of research that has sought to identify the kinematic patterns associated with many common soft tissue running injuries, including medial tibial stress syndrome (MTSS) (150), patellofemoral pain (PFP) (58, 192), iliotibial band syndrome (ITBS) (26, 59) and Achilles tendinopathy (AT) (121). Interestingly, many of these studies have reported similar kinematic patterns to be associated with different running injuries. For example, increased hip adduction has been associated with PFP (58, 192) and ITBS (26, 59) and increased hip internal rotation has been associated with PFP (188) and MTSS (150). Research has also suggested that due to the kinematic coupling between the femur, knee and foot, increased hip adduction or hip internal rotation may contribute to greater rearfoot eversion (132-134). Interestingly increased rearfoot eversion has been associated with injuries such as MTSS (55, 151) and Achilles tendinopathy (117, 121). This research suggests that there may be a number of similar kinematic patterns that could underlie multiple different soft tissue running injuries. It is possible that these patterns could lead to elevated stress on multiple anatomical structures leading to injury development at different areas. These kinematic patterns may represent global contributors to injury.

Recent research supports the idea of biomechanical parameters that could be considered global contributors to running injury. In a prospective study of 249 runners, Davis et al (53) reported that runners who went on to develop a range of different injuries, demonstrated significantly elevated vertical loading rates. While in a retrospective study which investigated runners with AT and MTSS, Becker et al (55) reported greater rearfoot eversion at late stance phase, to be a characteristic consistently associated with injury. Although these two studies provide preliminary evidence for the existence of global contributors to running injury, Davis et al (109) did not include kinematic data, while Becker et al (55) investigated only MTSS and AT. Therefore, further research is required to understand whether there are similar kinematic patterns that may underlie multiple different running injuries. This understanding would be invaluable to clinicians as it could be used as a basis for both screening techniques as well as preventative and rehabilitative programs.

5.1.1 Aim and Objectives

The aim of this current study was to investigate whether similar kinematic parameters are associated with multiple different common running related injuries. To achieve this aim, the objective of this study was to investigate whether there are differences in running kinematics between a large group of runners with common running injuries (ITBS, PFP, MTSS and AT) compared to a healthy control group. We hypothesised that the pooled group of injured runners would demonstrate greater contralateral pelvic drop, hip adduction and rearfoot eversion angles when compared to injury free controls. In order to ensure that differences observed were not the result of large effects in one of the injury subgroups, a secondary objective was to investigate whether kinematic differences observed between injured and healthy runners, differ between injury subgroups. We hypothesised that there would be no difference in kinematic parameters between injury subgroups.

5.2 Methods

5.2.1 Participants

A total of 108 runners were enrolled in this current study, including 72 injured runners (28 males, 44 females) and 36 healthy controls (15 males, 21 females) matched for age, height and weight (Table 25). The injured group contained subgroups of 18 runners with

PFP, ITBS, MTSS and AT (Table 26). These injuries were selected as they are cited as the most prevalent soft tissue overuse running injuries (12). An a priori sample size calculation was conducted using data from a previous study reporting kinematic differences between healthy and injured runners (192). Using g*power software, we calculated that we would need at least 98 people (65 injured) in order to detect an effect size of 0.75 with a power of 0.85 and a critical α of .01. Participants were recruited via poster advertisements at local running clubs and sports injury clinics. All participants provided written informed consent prior to participation and ethical approval was obtained via the local ethics committee.

*Table 25:Mean [SD] participant characteristics. *indicates statistical significance at p <.01.*

	Healthy (n = 36)	Injured (n = 72)
Sex (male/female)	15/21	28/44
Age (years)	33.2 (8.4)	34.8 (9.9)
Mass (kg)	60.8 (8.4)	63.4 (10.5)
Height (cm)	171.6 (7.3)	170.7 (8.6)
BMI (kg.m^{-2})	20.6 (1.8)	21.7 (2.7)
Miles run per week*	60.5 (23.2)*	21.2 (13.1)*

*Table 26:Mean [SD] injury subgroup characteristics. *indicates statistical significance at p <.01.*

	PFP (n = 18)	ITBS (n = 18)	MTSS (n = 18)	AT (n = 18)
Sex (male/female)	9/9	7/11	3/15	9/9
Age (years)	34.5 (9.4)	34.3 (7.9)	31.9 (9.7)	38.5 (11.7)
Mass (kg)	64.4 (9.6)	63.6 (11.2)	62.5 (10.1)	63.1 (11.8)
Height (cm)	173.5 (8.5)	170.6 (8.5)	167.3 (8.1)	171.6 (8.7)
BMI ($kg.m^{-2}$)	21.3 (1.9)	21.8 (3.3)	22.2 (2.3)	21.3 (2.0)
Miles run per week*	18.6 (6.9)	14.8 (5.8)	19.5 (12.2)	31.9 (17.6)*

5.2.1.1 Inclusion/ Exclusion Criteria

Inclusion and exclusion criteria were in accordance with that outlined in the methods Chapter, Section 3.2 and is summarised in the following section.

5.2.1.1.1 Injured Group

The injured group included individuals with a current diagnosis of either PFP, ITBS, MTSS or Achilles tendinopathy. Injury diagnosis was confirmed following a physical examination by a qualified physiotherapist in accordance with previously published diagnostic criteria for PFP (337), ITBS (228), MTSS (139) and Achilles tendinopathy (343) (Section 3.2.2, Appendix D). All participants reported being able to run up to 10 minutes before the onset of pain and maximal pain during running greater than 3/10 on a numerical rating scale (0 = no pain, 10 = worst possible pain). Additionally, all participants reported they were not currently receiving medical treatment for their injury and that their pain had caused a restriction to their running volume and/or frequency for a minimum of 3 months. Previous research has reported training factors such as increases in weekly training volume, to increase the risk of injury. This is likely due to a sudden excessive rise in acute tissue stress on the musculoskeletal system, resulting in insufficient time for adaptive changes (47). Therefore, in order to control for training errors as a cause of injury, participants were excluded if they reported an increase in weekly training volume of greater than 30% proceeding the onset of injury.

5.2.1.1.2 Control Group

Control participants were included if they reported running a minimum of 30 miles per week for the last 18 months with no reported injury. Participants were excluded if they reported any musculoskeletal ailment within the last 18 months that caused a restriction or cessation of running, or any need to seek advice from a health care professional. Additional exclusion criteria included previous history of overuse running injury, injury caused by another sport, previous spinal injury or lower limb surgery (Section 3.2.1.2).

5.2.2 Procedures

Kinematic data were collected from all participants whilst running on a treadmill at 3.2m/s wearing their own running shoes. After a 5 minute warm up period, 30 seconds of kinematic data were collected using a 12 camera Qualysis Oqus system (240Hz). A total of eight anatomical segments were tracked following a previously published protocol by the same authors, described in detail in Section 3.3, shown to have good to excellent repeatability (136, 201) (Chapter 4, Section 4.3). Segments included the thorax, pelvis and bilateral thigh, shank and foot segments. In addition, a further rearfoot segment was included using 3 non colinear markers attached to the heel of the participant's shoes. The foot segment was used to calculate sagittal plane ankle kinematics while the rearfoot segment was used to calculate frontal plane foot kinematics. Further details of the markers used to track each segment and the precise definition of the anatomical coordinate systems is provided in Section 3.3 and described in previous publications (136, 201, 235).

Raw kinematic data were low pass filtered at 10Hz. Intersegmental kinematics, along with the motions of the pelvis and thorax with respect to the laboratory system, were calculated using a six degrees of freedom model using the commercial software Visual 3D (C-Motion). Gait events were defined using a kinematic approach (363) and subsequently used to segment each kinematic signal into a minimum of 10 consecutive gait cycles. An ensemble average for each signal was created and selected kinematic parameters derived from the ensemble average curves. This latter processing was carried out using a custom Matlab script.

5.2.3 Data Analysis

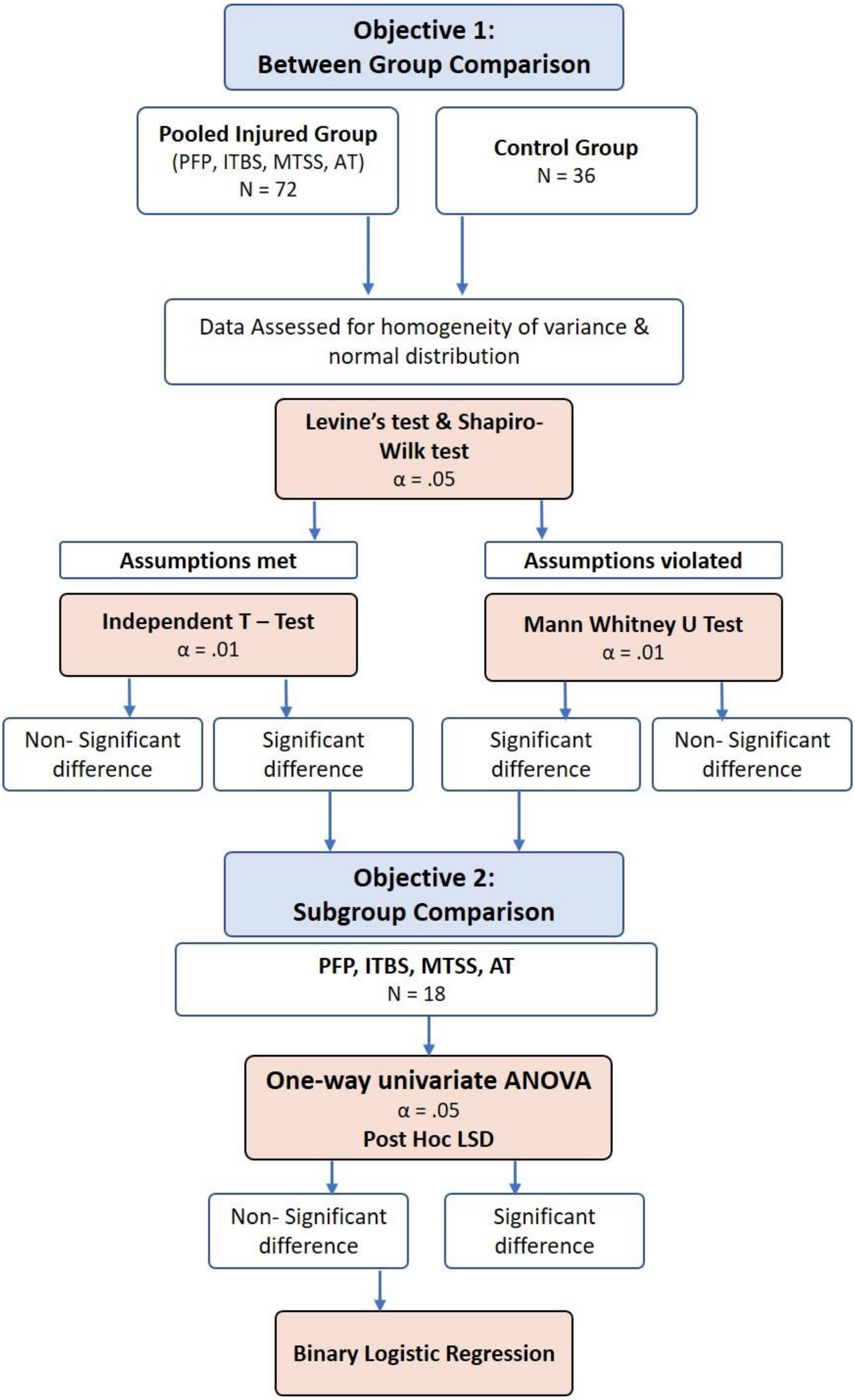
A range of kinematic parameters at both initial contact and during stance were selected for analysis. Parameters at initial contact included sagittal plane angles of the trunk, pelvis, hip, knee and ankle as well as frontal plane angles of the trunk and rearfoot. Peak angles during stance included sagittal and frontal plane angles of the trunk, pelvis, knee and ankle and rearfoot as well as transverse plane angles of the hip and knee. Parameters were selected based on those identified within the literature review as being associated with common running injuries, along with parameters with limited prior research (Section 2.1.7.2, Table 18). Peak angles during stance were defined as the maximum joint angle between initial contact and toe off. Foot strike patterns of each group were determined based on the kinematic waveforms of the ankle joint. Where the ankle demonstrated an immediate movement into plantarflexion, participants were classified as having a rearfoot strike, participants demonstrating immediate ankle dorsiflexion were classified as a forefoot strike. The injured leg was analysed from the injured runners, right or left leg was analysed at random from the healthy runners in order to match the total distribution of right and left legs in the injured group.

5.2.4 Statistical Analysis

Participant characteristics were analysed using independent t-tests for the healthy versus injured group comparisons and a one-way univariate ANOVA for the subgroup analysis (Table 25 & Table 26). Chi-squared tests were used to assess for differences in distribution of foot strike patterns and sex between the groups. In order to identify possible global contributors to running injury we used a two-phased statistical approach outlined in Figure 16. Firstly, data from the injured group were pooled and kinematic parameters compared with those of the control group. Before analysis, all kinematic parameters were assessed for homogeneity of variance and normal distribution using Levine's test ($p > .05$) and Shapiro-Wilk ($p > .05$). Where assumptions were met, between-group differences were assessed using an independent t-test. Where assumptions were not met, the Mann-Whitney U test for non-parametric test was used (375). Variables which failed to meet the assumption of normal distribution included peak angles of hip adduction, knee flexion and ankle dorsiflexion as well as angles at initial contact including trunk forward lean, trunk ipsilateral lean and ankle dorsiflexion. For control versus injured group comparisons a critical α of .01 was used.

Secondly, following identification of variables found to be significant different following the injured versus healthy comparison, we assessed for subgroup differences between the four injury subgroups. In order to be considered a global contributor to running injury, we required a kinematic parameter to be consistent across the different injury groups. This ensured that differences observed in the pooled injury data, were not the result of large effects in one of the injury subgroups. For this process, a one-way univariate ANOVA and post hoc Least Significant Difference (LSD) was used with a critical α of .05. The alpha level was set to .05 due to the smaller subgroup sample sizes, the smaller number of comparisons and to reduce the likelihood of type 2 error. Post hoc LSD tests were chosen as in contrast to alternative post hoc tests, as the LSD does not adjust for multiple comparisons, therefore maintaining statistical power and reducing the risk of type 2 error (375). This was deemed necessary in order to meet the aim of the subgroup comparisons, ensuring any subtle differences between subgroups were identified.

Figure 16: Flow chart of the statistical process and tests utilised to achieve the research objectives.



In addition to calculating statistical significance for group comparisons, we also calculated effect sizes. For t-test comparisons, we used Cohen's D and interpreted an effect size of 0.2, 0.5 and 0.8 as small, medium and large respectively (376). For the one-way univariate ANOVA comparisons, we used the Eta squared statistic (η^2 = SS between groups/ SS total) and interpreted effect sizes of 0.01, 0.09 and 0.25 as small, medium and large respectively (376).

Finally, a forward stepwise binary logistic regression analysis was conducted in order to determine which kinematic parameters could predict classification into either the injured or the healthy group. Parameters identified to be significantly different between control and injured groups were considered for the regression model. Variables were excluded from the regression model if they were found to demonstrate differences between injury subgroups (Figure 16). In order to control for potential confounding variables, sex and foot strike pattern were entered into the logistic regression model as covariates.

5.3 Results

5.3.1 Injured versus Healthy

The pooled data showed the injured runners to land with significantly more knee extension and ankle dorsiflexion (Table 27, Figure 19). At mid-stance, the injured runners were found to have significantly greater forward trunk lean, CPD (Figure 17A & Figure 18) and hip adduction (Table 28, Figure 17C & Figure 20). Large effect sizes of 1.37, 0.89 and 0.87 were observed for CPD, hip adduction and knee flexion at initial contact respectively (Table 27 & Table 28). Trunk forward lean at mid-stance and ankle dorsiflexion at initial contact demonstrated moderate effect sizes of 0.65 and 0.71 respectively (Table 27 & Table 28). Chi-squared tests found no significant difference in the distribution of foot strike patterns between the groups ($P = .332$) or sex ($P = .781$). In the healthy group there was a total of 17 forefoot and 19 rearfoot runners. In the Injured group there was a total of 27 forefoot and 45 rearfoot runners.

*Table 27: Kinematic parameters at initial contact. Mean [SD]. Data represents angle at initial contact in degrees. * indicates statistical significance at p <.01.*

		Control	Injured	P value	Effect Size
Trunk	Forward Lean	3.9 (2.9)	5.7 (3.9)	.03	0.52
	Ipsilateral Lean	2.5 (1.8)	3.1 (2.2)	.25	0.28
Pelvis	Anterior Tilt	5.9 (3.3)	7.0 (3.8)	.13	0.32
Knee	Flexion*	10.2 (4.8)	6.0 (4.9)	<.01*	0.87
Foot / Ankle	Ankle Dorsiflexion*	2.4 (6.5)	7.2 (6.9)	<.01*	0.71
	Rearfoot Inversion	8.7 (6.1)	6.2 (4.5)	.02	0.47

*Figure 17: A: Contralateral pelvic drop for healthy and injured groups. B: Contralateral pelvic drop for healthy and injury subgroups. C: Hip adduction for healthy and injured groups. D: Hip adduction for healthy and injury subgroups. PFP = patellofemoral pain, ITBS = iliotibial band syndrome, MTSS = medial tibial stress syndrome, AT = Achilles tendinopathy. Whiskers represent +/- 1SD. * indicates statistically significant differences for T-Tests (A & C) and subgroup ANOVA (B & D). Healthy group is shown in B & D for comparison purposes only.*

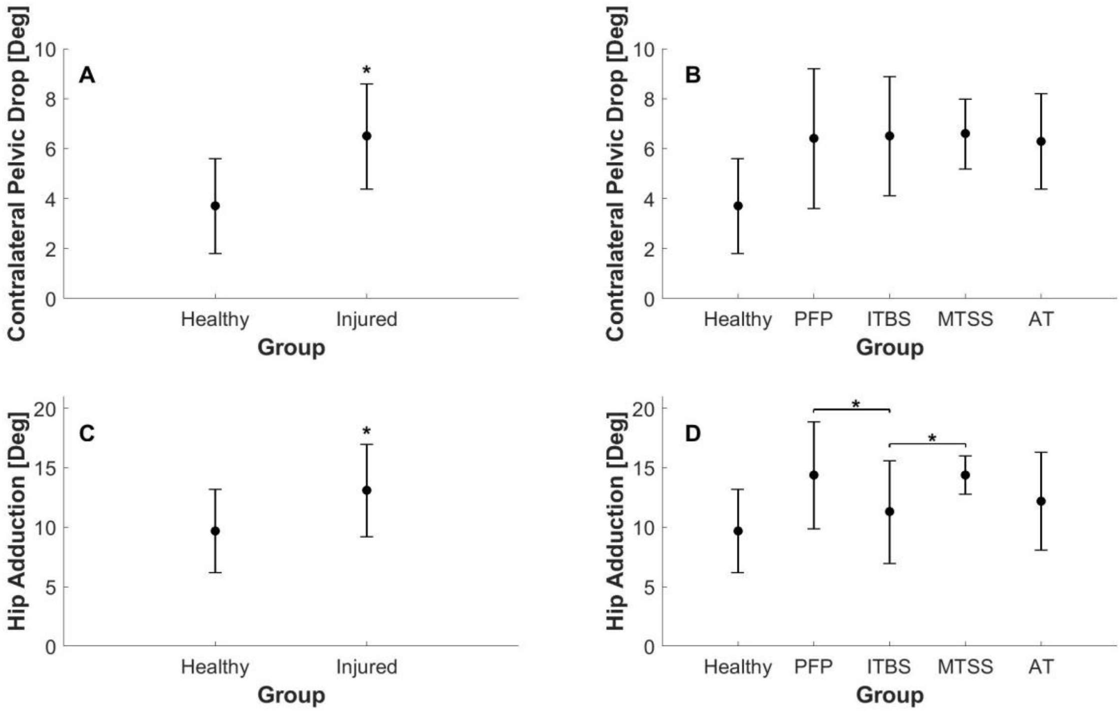


*Figure 18: Ensemble group average curve for frontal plane pelvis kinematics across the stance phase. Solid line represent group mean, shaded area represents 1SD. X-axis = percentage of stance phase. Y-axis = frontal plane pelvis angle in degrees, +ve values indicate contralateral pelvis drop, -ve values indicate contralateral pelvis elevation. *indicates statistically significant between groups.*

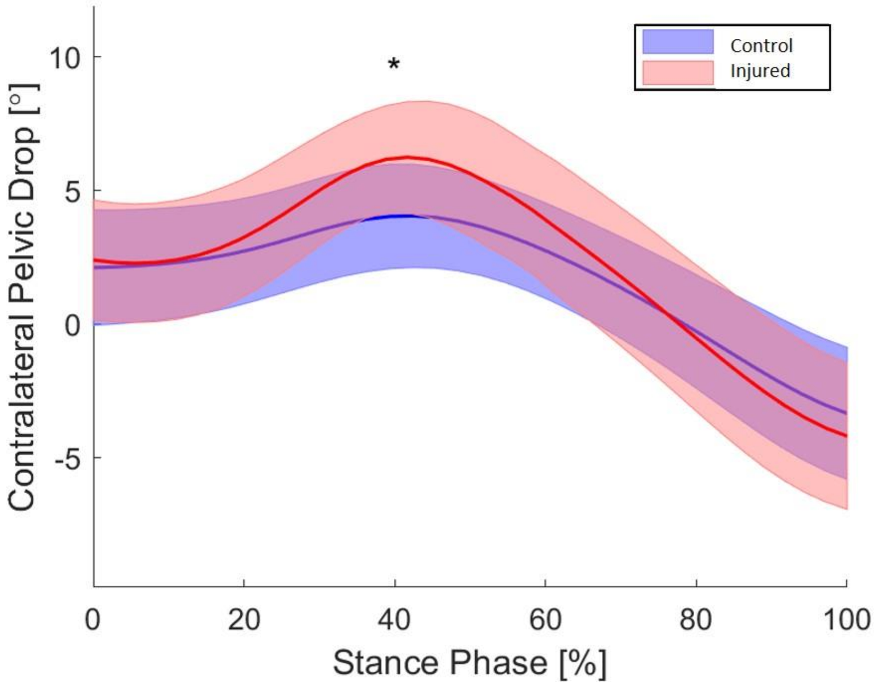


*Table 28: Peak kinematic angles during stance phase. Mean [SD]. Data represents the maximum joint angle between initial contact and toe off in degrees. * indicates statistical significance at p <.01.*

		Control	Injured	P value	Effect Size
Trunk	Forward Lean*	9.5 (2.9)	12.0 (4.9)	<.01*	0.65
	Ipsilateral Lean	3.6 (1.8)	4.3 (2.6)	.09	0.33
Pelvis	Anterior Tilt	5.0 (2.9)	5.7 (3.8)	.55	0.19
	Contralateral pelvic drop*	3.7 (1.9)	6.4 (2.1)	<.01*	1.37
Hip	Adduction*	9.7 (3.5)	13.0 (3.9)	<.01*	0.89
	Internal rotation	4.4 (6.8)	4.2 (8.0)	.87	0.03
Knee	Flexion	32.7 (4.9)	32.3 (5.0)	.56	0.09
	Adduction	-1.9 (3.1)	-2.0 (3.5)	.79	0.06
	External Rotation	6.7 (5.5)	7.1 (6.9)	.62	0.06
Foot / Ankle	Ankle Dorsiflexion	22.3 (2.9)	21.9 (4.3)	.96	0.09
	Rearfoot Eversion	2.6 (3.2)	4.0 (3.5)	.05	0.42

5.3.2 Injury Subgroups

The subgroup ANOVA analysis was conducted in order to identify if there were differences between injury subgroups for variables identified as being different between the pooled injured and healthy groups. This analysis found no differences for ankle dorsiflexion and knee flexion at initial contact (Table 29). Furthermore, there were no differences in peak trunk forward lean and CPD during mid-stance (Table 29), indicating these parameters were consistent across the injury subgroups. However, there was a significant difference between injury subgroups for peak hip adduction (Table 29). Post hoc LSD tests found the PFP (P <.01) and MTSS (P <.01) groups to have 3.1^{o} and 3.2^{o} more hip adduction than the ITBS group (Figure 17D).

*Table 29: Between injury subgroups ANOVA. Mean [SD] values are in degrees. * indicates statistical significance at p <.05.*

	PFP	ITBS	MTSS	AT	ANOVA	Effect Size Eta Squared (η^2)
Initial Contact						
Knee Flexion	5.5 (4.6)	6.6 (5.7)	4.7 (5.2)	7.4 (4.1)	.37	0.05
Ankle Dorsiflexion	10.6 (3.9)	7.1 (5.6)	5.5 (9.2)	5.6 (7.1)	.09	0.09
Mid Stance						
Trunk Forward Lean	11.9 (5.1)	14.3 (5.5)	10.9 (4.9)	11.3 (3.4)	.16	0.07
Contralateral Pelvic Drop	6.4 (2.8)	6.5 (2.4)	6.6 (1.4)	6.3 (1.9)	.99	0.002
Hip Adduction*	14.4 (4.5)	11.3 (4.3)	14.4 (1.6)	12.2 (4.1)	.03*	0.12

5.3.3 Logistic Regression

The final variables identified as global kinematic contributors included knee flexion and ankle dorsiflexion at initial contact as well as trunk forward lean and CPD at mid-stance. All four variables were entered into the logistic regression model with the inclusion of sex and foot strike pattern as potential confounding variables. The forward stepwise logistic regression model identified that CPD at mid-stance (OR = 1.87; 95% CI: 1.41, 2.49; $p < .01$) and knee flexion at initial contact (OR = 0.87; 95% CI: 0.78, 0.97; $p < .01$) were significant predictors of classification as either healthy or injured, explaining 47% of the variance in the data ($R^2 = .466$). The most important predictor variable was CPD, with an 80% increase in the odds of being classified injured for every 1° increase in pelvic drop. For knee flexion there was a 23% reduction in the odds of being classified injured for every 1° increase in knee flexion at initial contact. Sex ($P = .78$) and foot strike pattern ($P = .33$) did not have a significant effect upon the final model.

5.4 Discussion

This study identified a number of kinematic differences between the injured and healthy runners that were consistent across injury subgroups. In particular the injured runners were found to demonstrate significantly greater peak contralateral pelvic drop (CPD) and forward trunk lean, as well as a more extended knee and dorsiflexed ankle at initial contact (Table 27, Table 28 & Table 29) (Figure 19 & Figure 20). We found CPD to be the most important predictor variable when classifying runners as healthy or injured. Collectively, the observed kinematic patterns may increase tissue loads per stride during running, potentially contributing to the development of multiple different running related injuries.

Figure 19: Two-dimensional representation of forward trunk lean, knee flexion and ankle dorsiflexion angles at initial contact. A = injured runner, B = healthy runner.

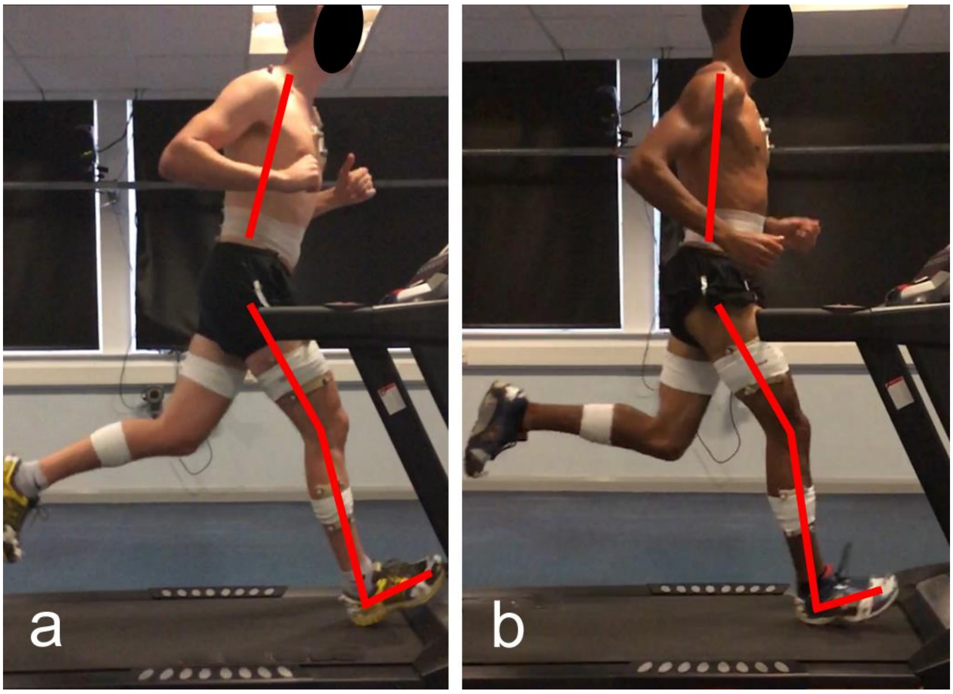
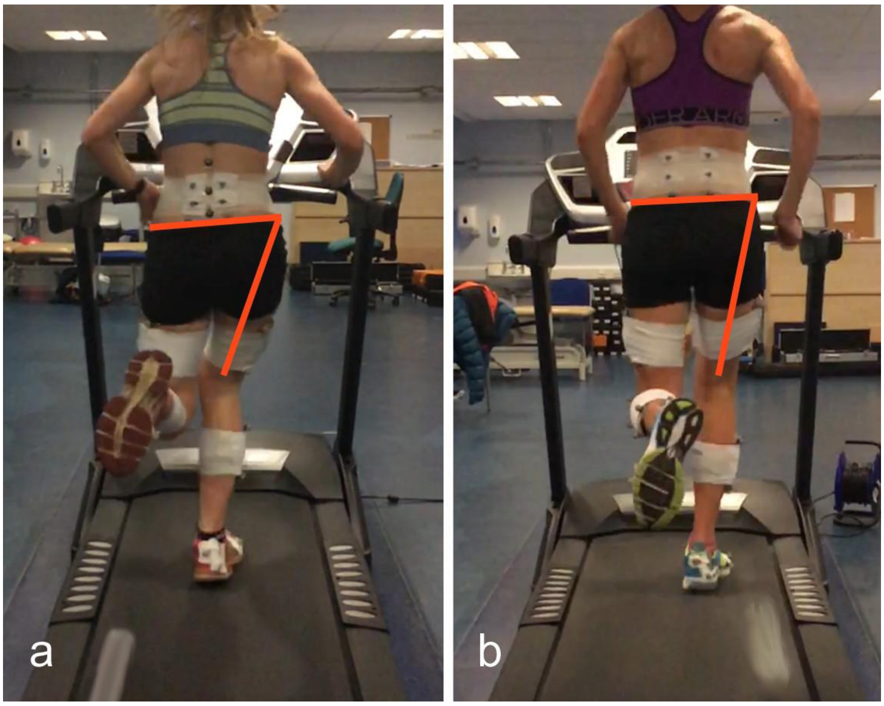


Figure 20: Two-dimensional representation of contralateral pelvic drop and hip adduction during mid-stance. A = injured runner, B = healthy runner.



5.4.1 Contralateral Pelvic Drop

Within Chapter 2, the literature review identified limited studies reporting the association between CPD and common running related injuries (Table 18). Two studies reported increased CPD amongst runners with MTSS (149, 150) and one study amongst runners with PFP (58), two reported no difference in CPD between ITBS runners and controls (240) and no prior study had investigated CPD amongst runners with AT. Interestingly, the current study found peak contralateral pelvic drop to be the kinematic parameter most strongly associated with running injury, present amongst multiple different running related injuries, including ITBS and Achilles tendinopathy (Figure 17B).

It is possible that CPD may increase the biomechanical loads placed on multiple different anatomical sites during each foot contact of a run. Contributing to the development of multiple different running related injuries via several different mechanisms. For example, Tateuchi et al (182) identified that increasing CPD resulted in an increase in iliotibial band tension at the lateral femoral condyle. This may influence ITBS development through increased strain rate (224) and increased compression between the ITB and lateral femoral condyle (219). At the same time, an increase in ITB tension will result in a lateral displacement of the patella (180). Lateral displacement of the patella will lead to a rise in patellofemoral joint stress, leading to PFP development (168), while at the lower limb, increased CPD will result in a medial shift in the ground reaction force relative to the knee joint centre (135, 136). This may alter the force distribution through the lower limb, leading to increased bending forces on the medial tibia (167) and potentially alter pressure distribution through the foot. This may contribute to the development of either MTSS or AT (98, 151).

One possible explanation for the increased CPD observed in the injured group could be due to reduced strength or neuromuscular function at the hip. Previous authors have reported delayed onset of gluteus medius and maximus in runners with PFP (138) and AT (137), while others have reported reduced hip abductor strength in runners with ITBS (377), PFP (188), AT (378) and MTSS (379). The hip abductors, in particular the gluteus medius, are thought to control frontal plane kinematics of the pelvis and hip (380). Therefore, it is conceivable that reduced strength or neuromuscular function of the gluteus medius would lead to an inability to stabilise the pelvis in the frontal plane, causing increased CPD. Further studies are now needed to investigate potential underlying mechanisms influencing CPD.

The finding of increased CPD amongst runners with ITBS is in contrast to the results of two prior studies reporting no difference between ITBS groups and controls (240). However, this may be explained by subtle differences between investigations. In particular, all participants in the current study were experiencing ongoing injury symptoms, whereas Foch & Milner (241) recruited runners who had been symptom free for a minimum of one month prior to testing. Through the inclusion of an asymptomatic population, it is possible that kinematic patterns associated with ITBS may have been resolved at the time of testing. In contrast, runners in the present study were experiencing ongoing injury symptoms, therefore it is possible that potential contributors to injury may have remained unresolved.

Results from a further study by Foch et al (227) support this possibility. When comparing runners with current ITBS to those with a prior history of ITBS, the group with a prior history of ITBS were found to demonstrate 1.9^{o} less CPD. Although the authors did not report the difference as statistically significant, the observed difference demonstrated a moderate effect size of 0.6 and exceeds the MDC value of 1.7^{o} reported for peak CPD in Chapter 4. Therefore, it is likely that this difference could be considered biomechanically meaningful, with the lack of statistical significance perhaps explained by the low participant numbers in each group (n = 9), limiting the statistical power to detect differences in pelvis kinematics. Consequently, these results support the premise that kinematic patterns driving ITBS may have been resolved at the time of testing in the study by Foch & Milner (241).

A further study by Foch et al (227) also reported no difference for peak CPD between runners with ITBS and controls. However, they did observe a significant 2.3^{o} difference in ipsilateral trunk lean, being greater amongst the current ITBS groups. As outlined within the literature review (Chapter 2, Section 2.1.7.2) ipsilateral trunk lean may serve as a compensatory pattern in order to minimise frontal plane pelvis displacement. In the present study, no difference was observed between groups for trunk ipsilateral flexion. This suggests participants in the present study did not appear to demonstrate compensatory kinematics to account for pelvis positioning and may explain the observed differences between studies.

5.4.2 Knee & Ankle Kinematics

We also found the injured runners to land with greater knee extension and ankle dorsiflexion (Table 27, Figure 19), which may influence tissue loads in a number of ways. Firstly, in knee extension the patella becomes vulnerable to lateral tilt and displacement which may influence patellofemoral contact areas and joint stress during early stance (381). Secondly, an extended knee and dorsiflexed ankle at initial contact is typically associated with a greater distance between the centre of mass and the foot at contact. Greater distance between the centre of mass and foot, as well as larger ankle dorsiflexion angles, have been associated with increased knee joint loading and breaking impulse (382). An extended knee at initial contact has also been reported to reduce the ability to attenuate impact forces during early stance (383). Collectively it seems plausible that the extended lower limb posture at initial contact may influence impact loading and knee joint loading during early stance.

5.4.3 Trunk Forward Lean

Increased trunk forward lean was a further kinematic pattern identified amongst the global injured group. Sagittal plane trunk kinematics are known to influence centre of mass positioning during running, with greater forward lean resulting in more anterior displacement of the centre of mass (136). This forward displacement of the centre of mass needs to be appropriately balanced in relation to foot positioning at contact, in order to maintain balance during running. If the centre of mass positioning is displaced too far anteriorly, then compensatory foot positioning may be necessary in order to maintain upright balance (292). Consequently, it is possible that the increased trunk lean observed amongst injured runners, may influence foot and lower limb posture at contact, resulting in the observed findings of increased knee extension and ankle dorsiflexion at initial contact.

One possible mechanism explaining the differences in forward trunk lean may be due to strength deficits around the gluteals and paraspinals. Previous studies have reported fatigue of the paraspinal and gluteal muscles to be associated with an increase in trunk forward lean during running (384) and drop landings (385). Therefore, reduced strength capacity of the gluteals and paraspinals may result in an inability to maintain an upright running posture amongst the injured runners.

Interestingly, no prior study has identified trunk forward lean to be associated with running related injuries (Table 18). In fact, some studies have suggested that increasing forward lean may represent a potential gait retraining strategy for injured runners (386). Using injury free populations, Teng & Powers (386) reported that increasing trunk lean serves to reduce both the knee extensor moment and patellofemoral joint stress during running (386). Subsequently, it has been suggested as a potential intervention strategy in the management of patellofemoral pain. However, based on the findings of the present study, it may be unwise to cue runners to increase forward lean during running. As failure to maintain the balance between trunk lean and foot position at initial contact, could lead to compensatory lower limb postures which may have negative consequences for further injury risk.

5.4.4 Magnitude of Difference

For the majority of kinematic parameters, the observed between-group differences exceeded the standard error of measurement identified within Chapter 4 (Section 4.3), highlighted in Table 30. The SEM is considered to represent the degree of error expected to occur within a measurement system. Considering the between-group differences were equivalent to, or exceeded the SEM values, the observed differences are likely to reflect true between-group differences in running kinematics. However, only peak CPD and hip adduction demonstrated a mean difference greater than the minimal detectable change (Table 30). Although this might not influence the interpretation of between-group differences, this could influence the future interpretation of intervention effects at an individual level. This is because averaging across a group is likely to reduce random (non-systematic) measurement variability. In contrast, at an individual level, changes in certain parameters, such as trunk lean and knee and ankle angles at initial contact, would need to result in large between day differences in order to be considered true intervention effects.

The minimal detectable change is considered to represent the degree of change representative of a true change, greater than that which could be explained by random error. However, the MDC does not necessarily represent the degree of change necessary for positive patient outcomes. Consequently, for those parameters with large MDCs but small between-group differences, the magnitude of change required to be clinically important for the individual, may be significantly smaller than the magnitude required to represent a true between day change in running kinematics. Subsequently, in many instances it may be difficult to evaluate whether interventions targeted towards certain kinematics, are responsible for changes to clinical symptoms.

Table 30: Visual comparison of the mean between-group difference for the pooled injured and control groups, compared to the SEM and MDC for kinematic testing procedures. SEM and MDC values are those presented in Chapter 4.

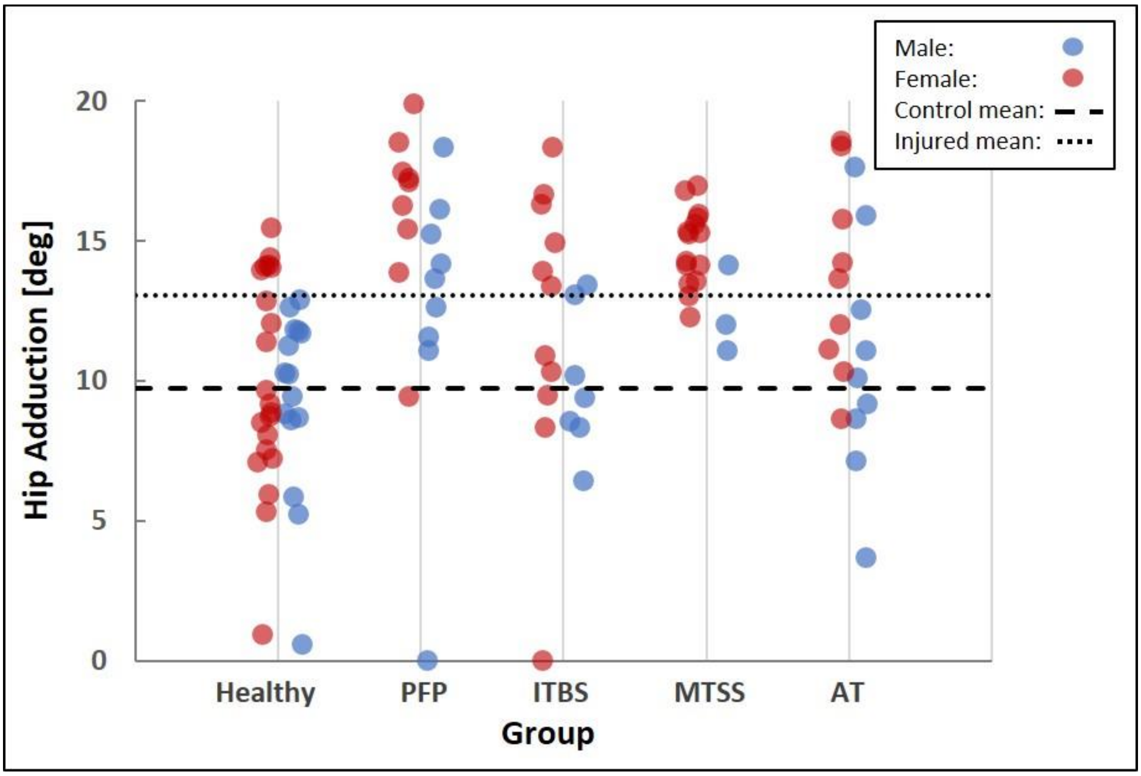
	Mean Difference (º)	SEM (º)	MDC (º)
Initial Contact			
Knee Flexion	4.2	2.5	6.9
Ankle Dorsiflexion	4.8	2.5	6.9
Mid Stance			
Trunk Forward Lean	2.5	2.6	7.1
Contralateral Pelvic Drop	2.7	0.6	1.7
Hip Adduction	3.3	0.7	1.8

5.4.5 Kinematic Subgroups

While hip adduction was found to be greater amongst the pooled injured group, the subgroup analysis revealed this parameter differed across the injury subgroups (Table 29, Figure 17C & Figure 17D). Specifically, we found hip adduction to be greater amongst subgroups of runners with PFP and MTSS compared to the ITBS subgroup (Figure 17D). This finding is in contrast to previous studies by Noehren et al (59) and Ferber et al (26) who reported increased hip adduction amongst runners with ITBS. One potential reason for the contrasting findings may be due to sex differences between studies. In the current study we included a mixed sex population, while Noehren et al (59) and Ferber et al (26), they only included female participants.

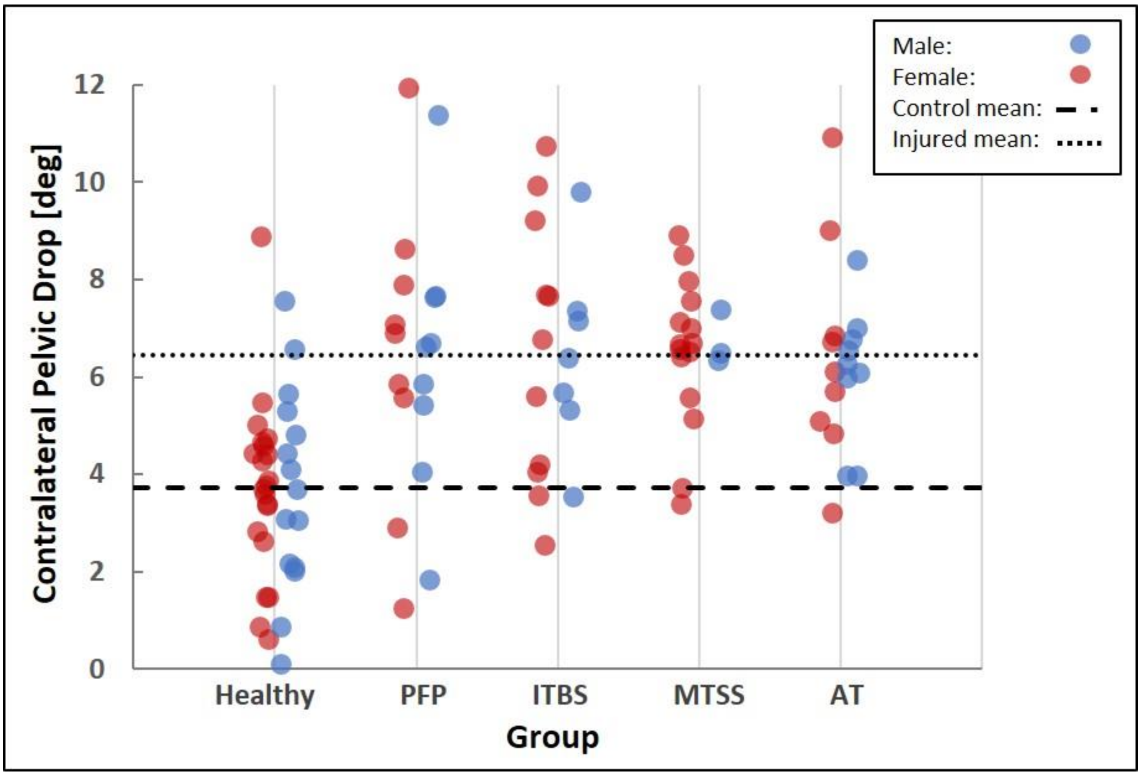
Hip adduction has been reported to be influenced by sex subgroups (58, 194, 238). Previous studies have reported that healthy female runners (238), as well as those with PFP (58, 194), tend to demonstrate greater peak hip adduction angles when compared to their male counterparts. Indeed, on inspection of the current data, hip adduction angles do appear to be influenced by sex. Specifically, within the current study, a higher proportion of male runners can be seen to demonstrate peak hip adduction angles below the average value for the pooled injury group (Figure 21). This is particularly true for the ITBS group, with 71% of male runners (5 out of 7) demonstrating peak hip adduction values lower than the average value of 13^{o} for injured runners, compared to only 45% of the female runners (5 out of 11). A proportion of these male runners also appear to demonstrate values lower than the average of 9.7^{o} for the control group (Figure 21). This suggests that although peak hip adduction may be an important kinematic risk factor for certain injuries, it is more frequently observed in female runners rather than male runners.

Figure 21: Individual hip adduction values for all subjects. Dashed line represents the mean hip adduction angle for the control group, dotted line the mean value for the pooled injured group.



In contrast to peak hip adduction, CPD appears less likely to be influenced by sex. On visual inspection of the individual data, there appears to be a similar distribution of male and female runners with CPD angles above and below the mean CPD values for the pooled injury group (Figure 22). This is further supported by results of the logistic regression model, as sex did not have a significant influence upon the results of the final model (Section 5.3.3). Therefore, while hip adduction may be more prevalent amongst female runners, CPD appears to be a kinematic pattern consistently observed across both sexes as well as injury subgroups.

Figure 22: Individual contralateral pelvic drop values for all subjects. Dashed line represents the mean contralateral pelvic drop angle for the control group, dotted line the mean value for the pooled injured group.



5.4.6 Role of Kinematics in Injury

Despite this study including multiple different running related injuries, similar kinematic patterns were observed across the injury subgroups. This raises an important question as to why certain musculoskeletal structures become injured rather than others. As presented in the introduction (Section 1.3.3), running kinematics are one factor influencing tissue loads per stride of a run. However, injury may only occur if the loads applied reach a cumulative load which exceeds the load capacity of specific tissues (28). This tissue specific load capacity is influenced by a variety of biological, psychological and sociocultural factors, influencing both the individual and tissue specific response to the biomechanical loads. The interaction between these factors, is likely to result in a tissue capacity which varies, both between individuals and tissues. This may explain why similar kinematic features could potentially influence multiple different running related injuries.

The results from the present study also highlight that some runners are able to remain injury free, despite demonstrating kinematic characteristics associated with injury. When inspecting individual data plots for the control group, a small proportion of runners demonstrated peak hip adduction and CPD values above the mean value for the injured group (Figure 21 & Figure 22). Five participants demonstrated peak hip adduction values (Figure 21) and three demonstrated CPD values greater than the average value for the pooled injured group (Figure 22). Considering these runners had reported no injury within the last 18 months and no prior history of common overuse injury, this highlights that kinematics alone are insufficient to explain injury development. This reflects the complex nature of running injury development that extends beyond singular risk factors, highlighting the need to consider the interaction between multiple risk factors if we are to fully understand injury aetiology and implement appropriate rehabilitation interventions.

One important consideration is the interaction between kinematics and training load exposure. According to Bertelsen's running injury framework (28), injury development is the result of the cumulate tissue load encountered during running, exceeding tissue load capacity. Although kinematics may increase tissue load per stride, without an exposure to external training load, the cumulative tissue load will remain relatively low and may be unlikely to exceed tissue capacity. Additionally, through the gradual progression of training load, it is possible that some runners may develop the tissue capacity required to tolerate the higher loads imposed by kinematic patterns (32, 268). This interaction between running kinematics and training load exposure has not previously been explored and therefore forms the motivation for Chapter 6, which aims to investigate whether kinematic parameters associated with common running injuries are associated with weekly training load exposure. Through achieving this aim, this may provide a theoretical understanding as to why some runners become injured as training volume increases, while others do not. It may also enhance our understanding of whether kinematics adaptations, if any, are required to attain regular high-volume training loads while remaining injury free.

5.4.7 Limitations

One limitation is the retrospective nature of the present study. Consequently, it is not possible to conclude if the observed kinematic patterns are the cause of injury, or the result of injury. Nevertheless, we ensured that all data were recorded before the onset of pain to minimise any possible effect of pain on the observed kinematic patterns. However, we cannot rule out the possibility that participants may have adapted their running kinematics in response to chronic injury or in apprehension of the acute onset of pain. Therefore, we acknowledge that future prospective studies are required to further investigate whether the kinematic patterns observed within the current study are the cause or effect of injury.

Another study limitation is the higher weekly mileage of the control group (Table 25). This was due to the stringent inclusion criteria required for the control group which may have introduced a selection bias to the current findings. It is important to note, the selection of a control group who have been injury free for 18 months, with no prior history of common overuse injuries, may have resulted in the selection of a specific heterogeneous population of runners. These runners may inherently possess characteristics, biomechanical or not, which allow them to remain injury free. However, we feel that this could be considered a strength of the current study, as previous research suggests running greater than 40 miles per week is a risk factor for developing injury (7). On average, our healthy control group were exceeding this threshold for more than 18 months prior to testing yet remained injury free. Therefore, we feel the control group may be representative of a healthy running gait in order to remain injury free at training loads exceeding the previously reported injury threshold.

There may also be behavioural differences between groups that has not been considered within the present study. For example, it is possible that some runners could have engaged in additional training activities, such as strength training, which was not accounted for. Subsequently, this may have influenced between-group differences in injury status or even running kinematics. However, there is currently inconclusive evidence to suggest that activities such as strength training, influence running kinematics (64, 65) or reduce the risk of running related injuries (387, 388). Therefore, we feel that despite possible differences in training routines, these may be unlikely to explain the biomechanical differences observed between groups. However, we acknowledge that future biomechanical studies should consider recording additional details of athlete training history and investigate the potential effects this may have upon running kinematics.

As all participants disclosed their injury status prior to kinematic data collection, it is possible that observer bias could have been introduced within the data collection and analysis procedures. However, in order to limit this possibility, all participants were screened against the clinical assessment and inclusion/ exclusion criteria prior to data collection and processing (Section 3.2). This ensured that the inclusion of participants was not influenced by prior knowledge of their kinematic patterns. Additionally, within the kinematic data collection protocol, pelvis kinematics were normalised to the standing trial, subsequently this would reduce any potential bias introduced through the positioning of retroreflective markers (Section 3.3.5). Therefore, we feel it is unlikely that observed bias would have influenced the present results.

It is also important to note that this study was limited to a select number of common soft tissue running injuries. Therefore, these results may not apply to other injuries such as plantar heel pain, stress fractures and muscle strains. Further research would be required in order to establish a link between the identified kinematic patterns and other running related injuries.

An additional limitation is the inclusion of a mixed sex population. As discussed earlier, hip adduction appeared to be influenced by gender subgroups within the data. This suggests that there may be sex specific kinematic patterns which could contribute to multiple different running injuries. Unfortunately, based on the power analysis conducted prior to this study (Section 5.2.1), the current study would have been underpowered to investigate the role of sex specific subgroups. However, the aim of the present study was not to identify sex specific running kinematics, but to investigate whether similar kinematic parameters are associated with multiple different common running related injuries. Considering peak CPD was found to be most strongly associated with common running injuries and appears to be consistent across both injury subgroups and gender, this parameter may represent a global kinematic contributor to injury, which can subsequently be targeted within the rehabilitation process.

5.4.8 Clinical Relevance

The findings from the present study may have a number of clinical implications. Firstly, all of the identified kinematic parameters can be easily visualised using two dimensional gait analysis methods (389-391) (Figure 19 & Figure 20). A number of recent publications have shown 2D assessments of CPD, hip adduction, trunk forward lean and sagittal plane knee and ankle angles to be highly correlated with 3D measurement systems and to demonstrate high intra and inter-tester reliability (389-391). Both the standard error of measurement and the minimal detectable difference for these parameters is also comparable to the results presented within the present thesis (Chapter 4) and smaller than the between-group differences observed within the present study. Specifically, Dingenen et al (392) reported a minimal detectable difference of 2.7^{o} for peak CPD and 2.8^{o} for peak hip adduction using 2D data collection methods. Therefore, it should be possible to use 2D measurement techniques to assess the biomechanical parameters which were associated with injury in this study.

Secondly, many of the identified global kinematic contributors to injury, can be modified through gait retraining. Within the literature review, several methods of gait retraining were found to effectively modify many of the observed kinematic patterns (Table 20). For example, CPD and hip adduction angles can be retrained using visual feedback (69) and increasing cadence (71), while knee and ankle angles are influenced by increasing cadence or modifying foot strike patterns (311). While the observed kinematic parameters may increase tissue loads per foot contact, gait retraining interventions, such as increasing running step rate, could modify kinematics and subsequently reduce tissue loads per foot contact. Building on this idea, Chapter 7 of this thesis aims to investigate whether a simple method of gait retraining can be used to improve biomechanics, clinical and functional outcomes amongst injured runners. This line of enquiry may provide preliminary evidence of the clinical effectiveness of a simple method of gait retraining targeted to the kinematic patterns observed in the present study.

5.5 Summary and Implications

This study identified several kinematic characteristics associated with common running injuries. In particular, we found injured runners to run with greater peak contralateral pelvic drop and trunk forward lean, as well as an extended knee and dorsiflexed ankle at initial contact. Contralateral pelvic drop appears to be the variable most strongly associated with common running related injuries. Based on theoretical models of running injury aetiology presented in the introduction (Section 1.3) and in order to achieve the overarching aims of this thesis (Section 1.3.7), Chapters 6 and 7 aim to expand on these findings and explore whether running kinematic are associated with training load exposure (Chapter 6) and whether gait retraining, targeted to the kinematic patterns observed in the present study, can be used to improve biomechanics, clinical and functional outcomes amongst injured runners (Chapter 7).

6 Chapter 6: Are running kinematics influenced by exposure to training loads?

In Chapter 5, several kinematic parameters were identified as being associated with common running injuries, with peak contralateral pelvic drop identified to be most strongly associated with common injuries. Based on injury causation models presented in the introduction, altered kinematic patterns were proposed to increase tissue load per stride of a run. When combined with an exposure to external training load, this may influence cumulative tissue loading and underlie injury development. Based on this premise, we may be less likely to see kinematic parameters associated with injury amongst runners who are able to attain high external training loads. However, currently there is limited evidence investigating whether kinematic parameters associated with common running injuries are associated with weekly training load exposure. Therefore, this Chapter presents a cross sectional study which aims to build upon the theoretical concepts presented in the introduction and explore whether the kinematic parameters, identified in the previous Chapter, are associated with external training loads.

6.1 Introduction

Training errors are frequently cited as the main cause of running related injuries (7, 23). Although several methods of monitoring training load exist (Section 2.3.2), from a running perspective, training loads are frequently measured using the external metric of weekly running volume. Perhaps influenced by the ease of measurement through the use of global positioning system (GPS) watches. However, currently there is conflicting evidence as to whether weekly training volume influences the risk of running related injuries (23, 42). Some studies have suggested running greater than 40 miles per week is a risk factor for running injury (43, 44), while others have suggested higher weekly training volumes are not associated with injury risk and may even be protective against injury (225, 228). This raises questions as to why some runners can attain high weekly training volumes and remain injury free, while others cannot.

Gabbett et al (281) proposed that high training loads alone are unlikely to be the cause of injury. Instead suggesting the rate of increase is likely to be the contributing factor; with sudden acute increases overwhelming the musculoskeletal system resulting in injury development (32, 281, 282). However, amongst running populations, there is conflicting evidence as to whether acute increases in training volume do increase the risk of injury development (42). When investigating week to week increases in training volume, several studies have reported no difference in injury rates amongst those increasing weekly training volume by less than 10%, compared to those increasing by 10% to 30% (24, 25, 26). This again raises questions as to why some runners become injured despite relatively small increases in weekly training loads, while others do not.

One possibility is that current training load measurements may not accurately reflect the cumulative tissue load encountered for a given run (30). According to Bertelsen's running injury framework (28), injury development is the result of cumulative tissue load exceeding tissue capacity. With cumulative tissue load defined as the sum of tissue specific load per stride and the frequency of load application. Current training load measurements only consider the frequency of load application and therefore as a sole metric, may not accurately reflect the cumulative tissue stress encountered by each individual. To accurately reflect the cumulative tissue loads it may be necessary to consider additional factors influencing tissue specific load per stride.

Within Chapter 5, several kinematic patterns were identified to be associated with common running related injuries. These parameters included increased peak contralateral pelvic drop, hip adduction, trunk flexion as well as reduced knee flexion and greater ankle dorsiflexion at initial contact. Interestingly peak contralateral pelvic drop was identified as the kinematic parameter most strongly associated with common running injuries. These kinematic parameters are thought to influence tissue load per stride, influencing the stress placed on musculoskeletal structures during each foot contact of a run. As kinematics may vary between individuals, when combined with an exposure to external training load, the cumulative tissue load encountered are also likely to demonstrate considerable between-subject variability. Consequently, this may cause some individuals to function closer to their tissue capacity, becoming injured at a much lower external training load or in response to relatively small increases in external load. Therefore, rather than injury risk being the result of the isolated variables of either training load or kinematics, it is perhaps the interaction between the two which is key to understanding the development of injury in a given individual.

It has been proposed that gradual progressive exposure to training may facilitate the development of physical qualities necessary to attain high training loads while reducing the risk of injury (32, 268). Indeed, some limited evidence does exist to suggest that runners may adapt aspects of their gait following exposure to a progressive training program (284). However, no current study has focused upon kinematic parameters commonly associated with running injuries. Based on the proposed interaction between training load exposure and running kinematics, it seems plausible to expect that runners who are able to attain regular high mileage running, while remaining injury free, could either adapt aspects of their running gait, or inherently possess kinematic patterns that reduce the stress placed on the musculoskeletal system. If this is the case, then this may have implications for load monitoring and load management amongst runners, offering a potential explanation as to why some runners become injured as training volume increases, while others do not.

6.1.1 Aim and Objective

The aim of this Chapter is to explore whether kinematic parameters associated with common running injuries are associated with weekly training load exposure. The specific objective was to investigate whether there is a difference, between groups of high and low-mileage runners, in the proportion of individuals who demonstrate kinematic patterns associated with injury. It was hypothesised that when compared to low-mileage runners, injury-free high-mileage runners would demonstrate a lower frequency of kinematic patterns shown (in the previous Chapter) to be associated with common running injuries.

6.2 Methods

In order to achieve the objective of this Chapter, a two-step process was used to assess for a between-group difference in the frequency of runners observed to demonstrate kinematic patterns associated with running injuries. Firstly, using the kinematic data presented in Chapter 5, a receiver operator curve analysis was conducted to determine cut off values for classification of injured runners using the kinematic parameter of peak contralateral pelvic drop. Second, the identified cut off value was applied to a new kinematic data set of high and low mileage runners in order to determine the proportion of individuals who demonstrate kinematic patterns associated with injury. Detailed methods are outlined in the following sections.

6.2.1 Participants

A total of 48 injury free runners were recruited for this study based on the inclusion criteria outlined in Section 3.2.1. Participants were separated into low (LM) (n = 24, 12 male, 12 female) and high-mileage (HM) groups (n = 24, 12 male, 12 female) based on their self-reported average weekly training volume. Average weekly training volume was selected as a measure of training load exposure as it provides an estimate of external training load exposure, representing the cumulative load encountered over a training week. Although it could be argued that training duration is a similar measure, measurement of weekly training volume appears to be a simple and quantifiable measure of external training load commonly used by runners, coaches and researchers (41, 272), with a large body of previous research having sought to investigate the association between weekly training volume and running related injuries. (43, 44, 45, 273 - 280).

Participants were assigned to the high-mileage group if they reported a weekly training volume equal to, or greater than 40 miles per week (64km) and the low-mileage group if they reported a weekly volume of less than 40 miles per week. The cut-off value for groups was based on previous literature suggesting an increased injury risk associated with weekly training volumes greater than 40 miles per week (43, 44, 275, 276). As such, it was hypothesised that runners who are able to exceed this training load without sustaining an injury, may be less likely to demonstrate kinematic patterns associated with common running injuries. Participant characteristics can be viewed in Table 31.

*Table 31: Mean [SD] participant characteristics. *indicates statistical significance at p <.01*

	Low Mileage	High Mileage	P Value
Age (years)*	40.5 (9.3)	30.7 (9.2)	<.01*
Mass (kg)	61.9 (7.7)	59.2 (8.9)	.24
Height (cm)	170.0 (9.6)	173.4 (8.9)	.21
Run Frequency (runs per week)*	3.6 (1.0)	7.8 (2.9)	<.01*
Weekly Run Volume (miles per week)*	22.8 (7.4)	59.8 (22.4)	<.01*

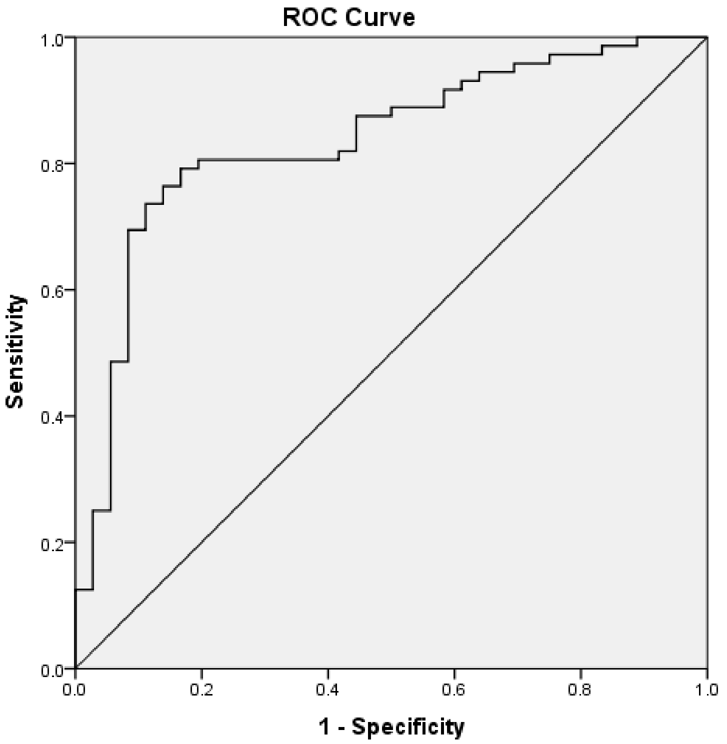
6.2.2 Kinematic data collection & processing

All kinematic data collection was completed and processed in accordance with methods outlined in Section 3.3.

6.2.3 Receiver Operator Curve (ROC) Analysis

Using the data presented in Chapter 5, Section 5.3.3, a ROC analysis was used to determine kinematic cut off values that would classify runners into injured groups and non-injured groups. A ROC analysis quantifies the accuracy of a test or measure, to discriminate between two outcome states (393). The ROC curve plots the test sensitivity and specificity across varying cut-off points with the area under the curve (394) providing a statistical interpretation as to the overall discriminative ability of the test (393). Chapter 5, Section 5.3.3 identified peak contralateral pelvic drop to be the kinematic parameter most strongly associated with common running injuries, therefore this parameter was selected for the ROC analysis and input as the test variable along with positive injury status as the state variable. The cut-off value for peak contralateral pelvic drop was determined based on observation of the curve points and coordinates in order to identify the coordinate point yielding the highest combined sensitivity and specificity (Figure 23). A cut off point of 4.82° was identified providing a sensitivity of 80.6% and a specificity of 80.6% when classifying runners into injured and non-injured groups (AUC = 0.835, p <.01, CI: 0.75 – 0.92).

Figure 23: Receiver operator cure (ROC) for peak contralateral pelvic drop using data collected in Chapter 5.



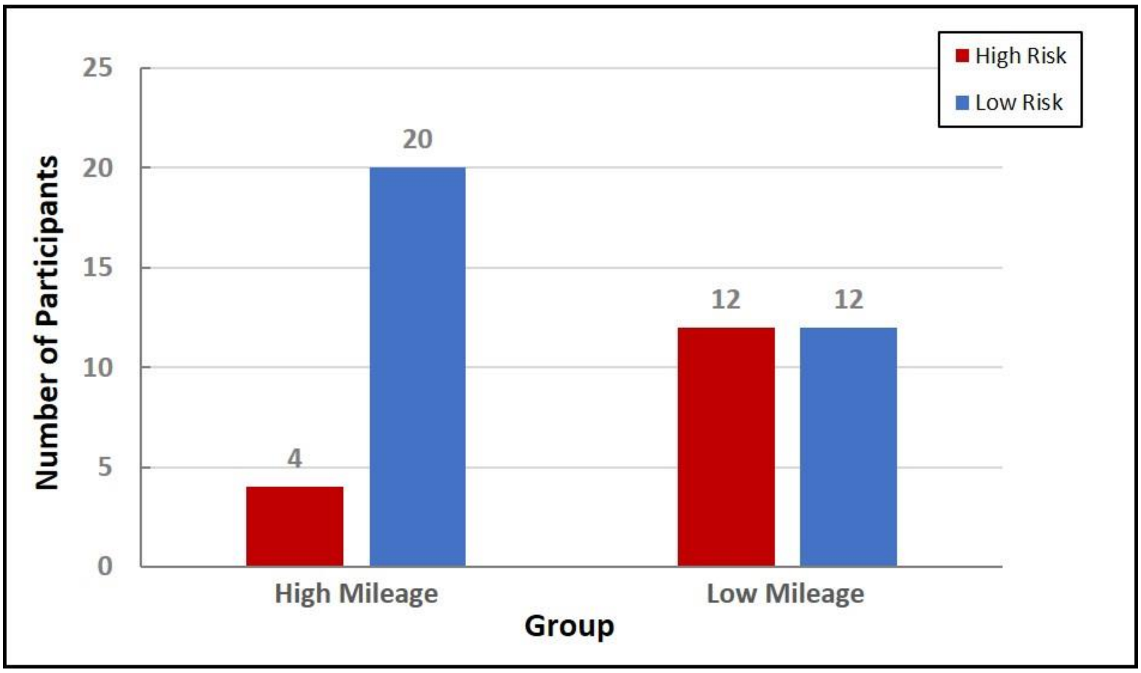
6.2.4 Statistical Analysis

Following identification of a cut off value, runners were classified into “high risk” and “low risk” groups based upon their peak contralateral pelvic drop angle. “High risk” was defined as a peak contralateral pelvic drop value equal to, or greater than 4.82° and “low risk” was defined as a value less than 4.82°. Pearson’s chi-squared test was used to assess for the difference in distribution of runners classified as demonstrating “high risk” or “low risk” CPD angles between the high-mileage and low-mileage runners. Descriptive statistics of minimum and maximum values were also calculated as well as the interquartile range to provide a measure of dispersion in the distribution of CPD angles within each group. The interquartile range was calculated as a measure of dispersion as it is less likely to be influenced by outliers within the data which may be the case when calculating the range (11).

6.3 Results

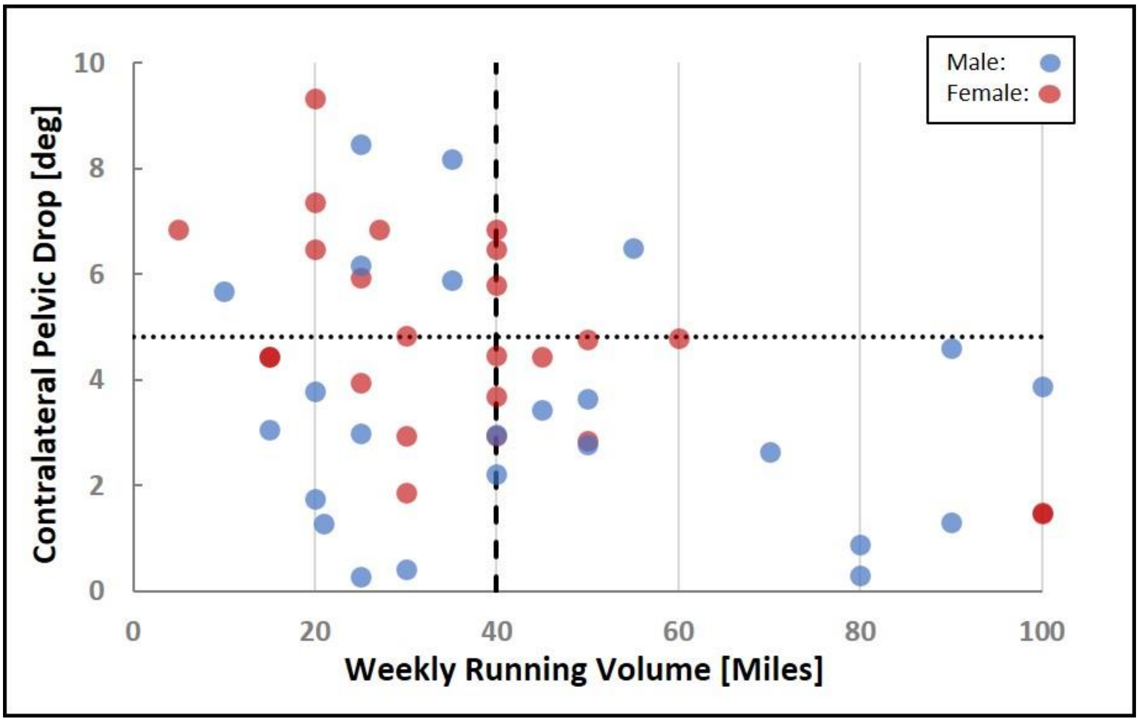
Pearson Chi-Square test identified a significant association between mileage groups and high or low risk CPD angles $\chi^2(1) = 0.6$, $p = .014$ (Phi and Cramer's V = .35). When compared to the low mileage group, the high mileage group contained a significantly lower frequency of runners with CPD angles, characteristic of running injury (Chapter 5) (Figure 24). Specifically, amongst the high mileage group, only 16.7% (n = 4) were classified as having “high risk” CPD angles, with 83.3% (n = 20) classified as “low risk”. In contrast, amongst the low-mileage group a total of 50% (n = 12) were classified as having “high risk” CPD angles and 50% classified as “low risk”.

Figure 24: Bar chart representing the number of runners classified as having "High Risk" and "Low Risk" peak contralateral pelvic drop (CPD) angles. X axis: high and low mileage groups, Y axis: number of runners. Blue bar represents "Low Risk" and red bar "High Risk" CPD angles.



The low mileage runners appeared to demonstrate greater within-group variability in peak CPD angles compared to the high mileage runners (Figure 25). Specifically, amongst the low-mileage runners, the minimum peak CPD angle observed was 0.3^{o}, with a maximum value of 9.3^{o} and an interquartile range of 3.6^{o}. In contrast, the high-mileage runners demonstrated a minimum value of 0.3°, a max of 6.9° and an interquartile range of 2.1°.

Figure 25: Scatter chart of individual plots for weekly running volume (x-axis) and peak contralateral pelvic drop (y-axis). Dashed vertical line represents the 40 mile per week cut off threshold used to separate runners into high or low mileage groups. Dotted horizontal line represents the CPD threshold of 4.82° used to classify runners as possessing "high risk" or "low risk" CPD angles.



6.4 Discussion

The objective of this study was to investigate whether there is a difference, between groups of high (HM) and low-mileage (LM) runners, in the proportion of individuals who demonstrate kinematic patterns associated with injury. In support of our hypothesis, high-mileage runners demonstrated a significantly lower frequency of individuals who exhibited CPD angles, characteristic of running injury (Chapter 5). The observed differences provide preliminary evidence for the existence of an interaction between kinematics and training load exposure.

6.4.1 Kinematics influencing the maximum workload potential

The observation of a lower frequency of HM runners with “high risk” CPD angles, may suggest that the loads induced through sub-optimal kinematics, are not sustainable at high volume training. Previous authors have suggested that each runner has a maximum workload potential, an external workload limit where structure specific capacity is exceeded (28, 30, 395). Within Chapter 5, peak CPD was identified to be associated with multiple different common running injuries and a rationale was presented to explain how this altered kinematic pattern could increase tissue loads throughout the musculoskeletal system during each stride of a run. When combined with a frequency of load application, such as during high external workloads, increased CPD could lead to cumulative loading which exceeds tissue capacity. Therefore, it is possible that runners who possess kinematic characteristics associated with common injuries (increased CPD), are more vulnerable to injury as training loads increase. This may explain the lower frequency of “high risk” CPD angles amongst the high-mileage group. Specifically, individuals who possess “high risk” kinematics may be unable to attain high external workloads without cumulative tissue loads exceeding tissue capacity.

In contrast, without an exposure to external training loads, kinematics may not be sufficient to cause injury development. This is highlighted by the findings of 50% of low mileage runners with “high risk” CPD angles. These angles were above the 4.82^{o} threshold, identified in Chapter 5 to be associated with common running injuries. Despite possessing kinematics associated with injury, these runners were injury free and reported no history of common overuse injuries. It is possible that the low training volume of these individuals means they have not exceeded a training load application sufficient to cause injury development. Therefore, these results highlight that injury causation is not as simple as possessing a risk factor or being exposed to an external training load. Instead, it is more likely to result from a complex interaction between multiple factors influencing the cumulative tissue loads encountered during running.

Based on the present results presented in this Section, it is possible that running kinematics could represent a potential effect-measure modifier, where the effect of training load upon injury incidence is modified by the kinematic features an individual possesses (396). Emerging evidence is beginning to highlight the role of effect measure modifiers in running injury aetiology. In a 12-week prospective study of recreational runners, van der Worp et al (128) reported that total weekly running volume was not associated with an increased risk of running related injury. However, when combined with having a previous history of injury, running greater than 30km per week significantly increased the risk of sustaining a running related injury. Similarly, Malisoux et al (396) reported that for a given training exposure, the risk of sustaining a running related injury is significantly greater amongst runners with a previous history of injury.

In the context of the present findings, baseline running kinematics could provide one explanation for the conflicting literature regarding the influence of training loads upon running injury development (23, 42). This is because the influence of training load may be modified by the baseline kinematics of an individual. With those who possess kinematic characteristics which increase tissue load per stride, more vulnerable to injury development as the frequency of external load application increases. Therefore, future studies should consider kinematics as a potential effect measure modifier. With this approach, stratifying groups of runners based on their baseline kinematics would allow for further understanding of whether kinematics do influence the training loads attainable by runners.

6.4.2 Kinematic adaptations to external training loads

An alternative explanation for the current findings is that some runners may adapt aspects of their gait in response to elevated training loads. Several authors have proposed that progressive exposure to training loads may facilitate the development of physical qualities necessary to attain high training loads, while reducing the risk of injury (32, 268, 397). Although limited, some evidence does support this idea. Following a 10-week beginner running program, Moore et al, (284) observed pre and post program changes in several kinematic parameters, including a reduction in peak rearfoot eversion velocity. The authors hypothesised that the observed changes, may be adaptions to improve running economy and reduce the risk of musculoskeletal injury associated with certain kinematic parameters. Therefore, it is possible that the observed between group differences may represent adaptations to high training loads, with these adaptations occurring in order to reduce the risk of injury associated with kinematic patterns such as CPD.

It has also been proposed that progressive exposure to training loads may facilitate tissue adaptation, improving capacity to withstand load application (32, 282, 398). In the context of kinematics, this would imply that gradual exposure to external training loads, may facilitate the development of tissue capacity to tolerate the loads imposed through sub-optimal kinematic parameters, such as increased peak CPD. The data presented in this current Chapter could be used to either support or refute this possibility. In support of this possibility, 4 out of the 24 high-mileage runners were observed to possess CPD angles which could be considered “high risk” (Figure 24). Additionally, within Chapter 5, three runners within the control group demonstrated CPD values greater than the average value for the pooled injured group (Figure 22). This suggests that some individuals may be able to withstand the tissue loads induced through sub-optimal running kinematics yet remain injury free at high external training loads.

Nevertheless, at a group level, the idea that tissues adapt to tolerate the loads induced through kinematics is not supported by the current data. Although 4 high-mileage runners demonstrated “high risk” CPD angles, the frequency of “high risk” kinematics was significantly less than that of the low-mileage group. Furthermore, if the cut off threshold for external training volume was increased to 41 miles per week, then the number of high mileage runners with “high risk” kinematics would reduce to only 1 participant (Figure 25). This suggests that, as the external training demands increase, less runners appear to possess kinematic characteristics similar to those associated with common running injuries. Therefore, we feel that the present results may support the idea of a possible biomechanical Darwinism amongst runners. Specifically, in order to attain regular high mileage running and remain injury free, runners must either adapt aspects of their gait, or inherently possess kinematic features that minimise the stress placed upon the musculoskeletal system.

6.4.3 Implications for training load management

The findings of the present study may have several implications for load management and load monitoring. Current methods of load monitoring tend to focus upon the physiological response to load through the use of internal metrics, or the quantification of external loads via metrics monitoring the training "dose", such as volume or duration of running. Although these methods are useful in determining the external exposure and individual response to training, they do not consider factors influencing tissue specific loads per stride, such as running kinematics. Consequently, current metrics may not accurately reflect the cumulative tissue loads encountered during running.

It is possible that, for a given external training load, the tissue loads imposed through sub-optimal kinematics may result in a cumulative load that is significantly greater than that of a runner who does not possess sub-optimal kinematics. As suggested above, this may provide one explanation as to why some runners can attain high training volumes and remain injury free, while others cannot. Therefore, for individuals who possess sub-optimal running kinematics, a more cautious approach to training load progression may be necessary. For such people, regular monitoring of individual response to training demands and smaller week-to-week increases in training volume, may allow for the opportunity for kinematic adaptations and prevent overwhelming the musculoskeletal system.

Finally, amongst individuals with sub-optimal kinematics, gait retraining interventions may help to reduce the risk of injury and facilitate higher attainable workloads. Some preliminary evidence suggests that baseline gait retraining may reduce the risk of future running related injuries (399). In a study of 320 novice runners randomised into a gait retraining and control group, at one year follow up the gait retraining group demonstrated a significantly lower injury incidence compared to the control group. Therefore, it is possible that gait retraining interventions which effectively target kinematics associated with common running injuries, may help to reduce tissue loads per stride, allowing for higher attainable workloads. Additionally, amongst runners who have exceeded their load capacity and possess sub-optimal kinematics, gait retraining may improve kinematics, reduce tissue loads and improve clinical outcomes amongst injured runners. Building on this concept, Chapter 7 of this thesis aims to explore this further by investigating whether a simple method of gait retraining can be used to improve biomechanics, clinical and functional outcomes amongst injured runners who possess sub-optimal kinematics.

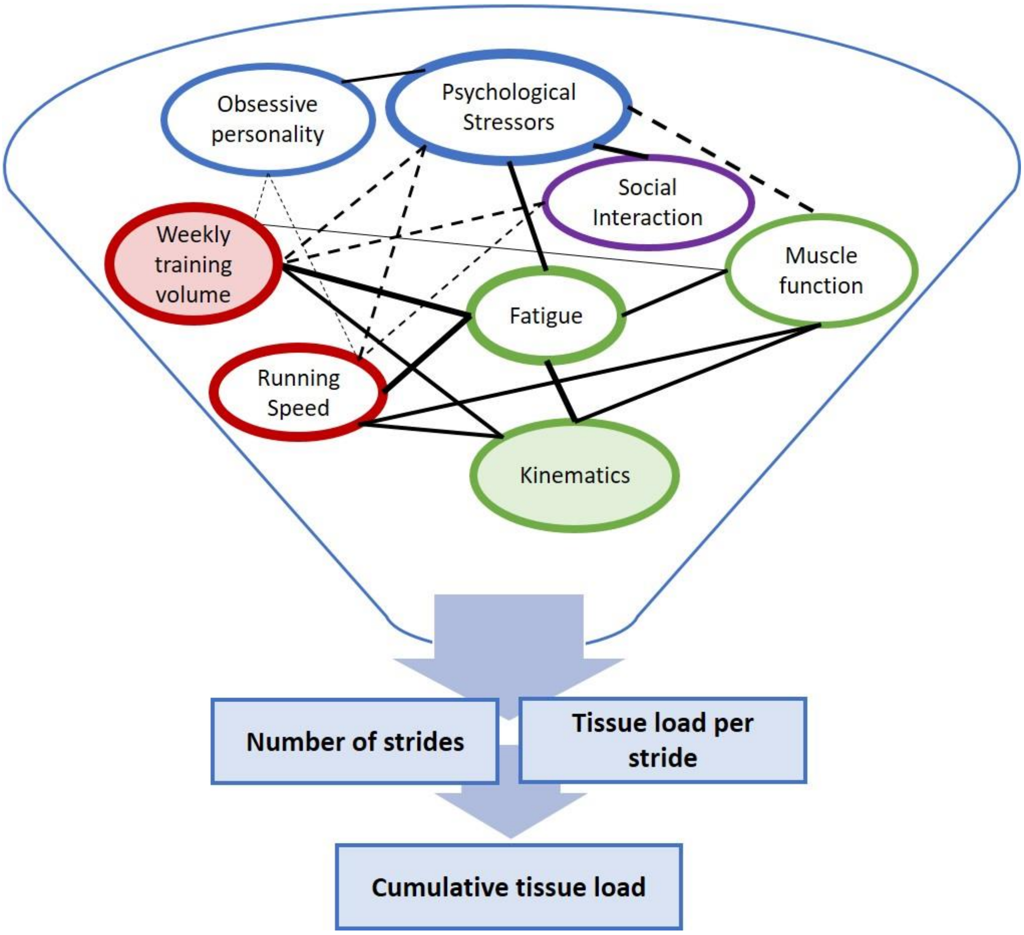
6.4.4 Limitations

There are several limitations to the present study that should be acknowledged. Firstly, it should be noted that there was a between group difference in age, with the high mileage group on average 9.8 years younger than the low mileage group. However, no prior study has reported frontal plane pelvis kinematics to be influenced by age (400-403). Additionally, a systematic review by van der Worp et al (374) concluded that there is limited evidence to suggest age is a significant risk factor for running related injuries. Therefore, the age difference is unlikely to be large enough to explain the observed findings within the present study.

A second limitation is the cross-sectional nature of this study, which means the results cannot be used to predict future injury risk amongst the low-mileage runners, nor can it predict high-mileage runners will not develop injury. However, it does raise an interesting question as to why high-mileage runners appear to demonstrate a lower frequency of kinematic patterns shown (in the previous Chapter) to be associated with common running injuries, and whether the higher frequency of low-mileage runners with “high risk” kinematics could influence their future injury risk. Future studies should consider exploring the role of kinematics as a potential effect-measure modifier. Prospective cohort studies could consider investigating whether those who possess sub-optimal kinematics at baseline, are more vulnerable to injury as training loads increase.

Finally, it is important to note that although the current study highlights potential interaction effects between two single variables, interactions will likely extend beyond the parameters studied. Bittencourt et al (19) previously described injury aetiology to be influenced by the “web of determinants”. The complex interaction between multiple psychological, biological and sociocultural factors which adapt over time and can mediate the effect imposed by additional factors. Similar to injury causation, the constructs of tissue load and the frequency of load application are also likely to be influenced by this complex web of determinants. With factors interacting to influence not only the outcome of cumulative tissue load, but the magnitude of effect of the singular components (see Figure 26 for an illustrative example). For example, running kinematics may be influenced by fatigue, with fatigue influenced by a variety of training and psychological factors (Figure 26). Similarly, social influences may influence training behaviours either directly, or indirectly through imposed psychological stressors. This in turn, may influence fatigue, running kinematics and the resulting frequency of load application and tissue loads per stride. Therefore, from both a clinical and research perspective, it is important to not only consider the individual components influencing tissue load, load exposure and tissue capacity, but to also acknowledge the wider interactions between multiple factors which may influence the singular components and the wider outcomes.

Figure 26: Conceptual diagram representing the interaction between multiple individual variables and their influence upon cumulative tissue load. Note: this is not an exhaustive list of all variables which may influence tissue load and load application.
The interaction between individual variables is depicted as the web of determinants as described by Bettencourt et al (19), with the diagram highlighting their influence upon components of the framework for running injury aetiology described by Bertelsen et al (28). Blue circles = psychological factors, purple = sociocultural, green = biological, red = training parameters. Lines represent the strength of interaction between variables with thicker lines indicating a stronger interaction and dotted lines indicating weaker interaction. Thicker circles represent more interactions between variables. Highlighted circles are those included within the present study.



6.5 Summary and Implications

The results from this study highlight an association between training load exposure and running kinematics. In particular, when compared to low-mileage runners, injury-free high-mileage runners demonstrated a significantly lower frequency of kinematic patterns similar to those associated with common running injuries. It is possible that the tissue loads imposed through contralateral pelvic drop, could influence the maximal workloads attainable. With runners either adapting aspects of their kinematics or becoming injured as training loads increase. This may provide a theoretical explanation as to why some runners develop an injury as training loads increase while others do not.

In such instances, gait retraining interventions specifically designed to improve kinematic patterns may prove effective in reducing the cumulative tissue loads encountered and improve clinical outcomes amongst injured runners. Building on this idea, Chapter 7 aims to investigate whether a simple method of gait retraining can be used to improve biomechanics, clinical and functional outcomes amongst injured runners.

7 Chapter 7: A 10% increase in step rate improves running kinematics and clinical outcomes in runners with patellofemoral pain at 4 weeks and 3 months follow up.

This Chapter is a case-series study which aims to investigate whether a simple method of gait retraining can be used to improve kinematics, clinical and functional outcomes amongst injured runners. Chapter 5 identified kinematic parameters associated with common running injuries and Chapter 6 discussed how, when combined with an exposure to external training volumes, the cumulative tissue load encountered could result in injury development. This Chapter builds upon on these findings and the concepts presented within the introduction, targeting a clinical intervention to the kinematic parameters identified within Chapter 5. The results provide preliminary evidence for the clinical effectiveness of a simple method of gait retraining amongst injured runners. The method employed can be easily integrated into clinical practise without the need for close clinical supervision and therefore offers a practical retraining method for clinicians. The repeatability results reported in Chapter 4 are used to aid interpretation of the current findings.

Following peer review, the results of this Chapter have been published within the American Journal of Sports Medicine (Appendix G). The following account includes an extended discussion of the published work:

Bramah, C., Preece, S, J., Gill, N., Herrington, L. (2019) A 10% increase in step rate improves running kinematics and clinical outcomes in runners with patellofemoral pain at 4 weeks and 3 months. *American journal of Sports Medicine.* 47 (14), 3406 – 3413.

7.1 Introduction

Recreational running is an increasingly popular method of physical activity with participation rates growing annually. Although running offers several health benefits, it also poses a considerable risk of injury to the musculoskeletal system. Overall injury incidence rates are reported to range between 19 and 78% amongst recreational runners (7) with reoccurrence rates in 20% to 70% of all cases (17). Of all running injuries, patellofemoral pain (PFP) is considered the most common running related knee injury (13) with incidence and prevalence rates as high as 20.8% and 22.7% respectively (14).

Patellofemoral pain (PFP) is known to have a multifactorial aetiology with aberrant running mechanics identified as one risk factor (57, 185, 404). Runners with PFP have been reported to demonstrate increased hip adduction (57, 58, 192), hip internal rotation (192) and contralateral pelvic drop (58) when compared to injury free controls. Within Chapter 5 contralateral pelvic drop and hip adduction were both identified to be associated with the PFP subgroup of runners. It is thought that kinematic patterns such as these may increase tissue load per stride of a run by influencing lateral tracking of the patella, leading to a rise in patellofemoral joint stress (168). When exposed to repeat loading cycles during running, this may result in damage to the underlying chondral surface, stress within the subchondral bone and excitation of nociceptors leading to pain and injury (169).

Gait retraining is a clinical intervention that targets running kinematics within the rehabilitation of PFP. Based on the literature review findings reported in Chapter 2 (Section 2.4.7), current evidence has shown improvements in kinematics and clinical outcomes following mirror retraining (69), real time feedback (70) and transitioning to a forefoot contact (73, 294) (Table 20). However, there are several limitations to current gait retraining methods. Mirror and real time feedback are restricted to clinical and laboratory settings limiting their practical applicability, while transitioning to a forefoot strike has been shown to increase Achilles tendon and ankle joint loading, which may increase the risk of lower limb injury (320). Furthermore, these studies often utilise a faded feedback design consisting of 8 sessions over a 2-week period, requiring close clinical supervision. Therefore, there is a need for gait retraining methods that can be easily integrated outside of a laboratory setting while providing positive clinical and biomechanical outcomes.

Increasing step rate may be one method of gait retraining that could be integrated outside of a laboratory setting. Through the use of Global Positioning System (GPS) “smart” watches and portable mobile metronome applications, runners may be able to self-retrain and monitor their step rate without the need for close clinical supervision (328, 334, 405). Currently only three studies have investigated the effects of increasing step rate amongst runners with PFP (255, 267, 318). Neal et al (255) reported improved frontal plane hip and pelvis kinematics along with reductions in pain, but did not investigate whether improvements were maintained beyond the 6 week follow up period. Escuilier et al (318) reported gait retraining to be no more effective than education on load management, and dos Santos et al (267) reported minimal changes in pain following a two week retraining period. Furthermore, both Escuiler et al (318) and Dos Santos et al (267) did not report any change in frontal plane hip and pelvis kinematics following the retraining period. Therefore, questions remain regarding the clinical effectiveness of increasing step rate and whether step rate retraining results in long term kinematic adaptations amongst runners with PFP.

7.1.1 Aim and Objective

The aim of this current study was to investigate whether a simple method of gait retraining can be used to improve biomechanics and improve clinical and functional outcomes amongst injured runners. In order to achieve this aim, the specific objective was to investigate whether a 10% increase in running step rate influences frontal plane kinematics of the hip and pelvis, as well as clinical outcomes in runners with PFP. It was hypothesised that a 10% increase in step rate will result in significant reductions in frontal plane hip and pelvis kinematics, improvements in clinical outcomes and function.

A secondary objective was to investigate whether runners can self-administer a 10% increase in step rate using an audible metronome and a GPS smart watch and whether these changes can be maintained at short term and long term follow up. If runners can self-administer retraining sessions this may prove to be a simple clinical intervention which can easily be integrated outside of the laboratory and clinical setting. It was hypothesised that runners will increase their step rate by 10% at short term follow up which will be maintained at long term follow up.

7.2 Methods

7.2.1 Participants:

Participants were recruited through advertisements at local sports injury clinics, running clubs and those attending a University based running clinic. Ethical approval for the study was obtained via the local ethics committee and all participants provided written informed consent prior to participation. This study was registered as a clinical trial (ClinicalTrials.gov, registration number NCT03067545) with enrolment for the trial between March 2017 and December 2018. An a priori sample size calculation was conducted using data from a previous gait retraining study identifying a 2.3^{o} reduction in contralateral pelvic drop post retraining with an effect size of 1.09 (70). Using G Power software, we calculated that 12 participants would be required to detect an effect size of 1.25 with a power of 0.8 and an adjusted critical alpha of .016. This calculation was based on the use of paired tests to detect differences in peak contralateral pelvic drop which were similar to changes observed in previous studies following gait retraining (2.3^{o}) (70) and also of similar magnitude to differences observed between injured and healthy runners in Chapter 5 (2.7^{o}) (406).

7.2.2 Inclusion/ Exclusion criteria

All participants were required to own a GPS smartwatch or running watch capable of monitoring step rate. Participants were included in the gait retraining intervention based on a three-stage assessment process. First, a subjective assessment and clinical examination were used to confirm the presence of PFP. Once the diagnosis of PFP was confirmed a 3D gait analysis was conducted to confirm the presence of hip and pelvis kinematics in a range identified to be associated with injury in Chapter 5. To ensure injury diagnosis met the consensus definition of a running related injury (335), participants had to report insidious onset of anterior knee pain during running lasting for a minimum of 3 months causing a self-reported restriction to either their running volume or duration. Participants were required to be running a minimum of twice per week with their worst pain rated a minimum of 3 out of 10 on a numerical rating scale (NRS) for pain (0 = no pain, 10 = worst possible pain). Pain must also be reproduced by one or more of the additional activities of either squatting, kneeling, prolonged sitting, ascending or descending stairs. Participants were excluded if they reported having any known medical condition, prior musculoskeletal surgery, neurological impairment, diagnosed knee osteoarthritis, structural deformity of the knee, the onset knee pain to be caused by trauma or any other sporting activity, had ceased running or were receiving additional treatment outside of the study. To control for training errors as a potential underlying cause of injury, participants were also excluded if they reported the onset of symptoms to occur following an increase in their weekly training volume equal to, or greater than 30% (47).

Following the subjective assessment, participants were invited to a clinical examination led by the lead clinician to confirm the diagnosis of PFP in accordance with previously published diagnostic criteria (Section 3.2.3) (336). Specifically, for inclusion to the study pain must be retropatella or peripatellar in nature and reproduced on squatting with the exclusion of any patella instability, ligamentous or meniscal injury (336). Pain on squatting has been shown to have a sensitivity of 91% and a negative predictive value of 74% suggesting this test to be the best available test for PFP (336, 338). A combination of additional, but non-essential, clinical tests was used to further increase the diagnostic accuracy of PFP (339). Tests included patella compression, patella apprehension, pain on palpation of the lateral patella facet and pain on resisted quadricep contraction in 30^{o} knee flexion (338, 339). These tests are known to have low sensitivity and specificity when used in isolation and were not used as a sole diagnostic criterion for PFP (338, 339).

Once the diagnosis of PFP was confirmed, each participant underwent an initial 3D gait analysis as outlined below and completed clinical outcome measures to monitor pain and functional improvements. Clinical outcome measures included the Lower Extremity Functional Scale (LEFS) (Appendix H), previously validated for use in PFP (326), as well as self-reported worst pain experienced in the past week using the NRS (0 = no pain, 10 = worst possible pain). These outcome measures were selected to allow for comparison of the current findings, to those of previous gait retraining studies utilising LEFS (69, 70, 267) and NRS for worst pain in the past week (255, 267, 318). Additional outcomes monitored were self-reported longest distance run pain free and total weekly running volume.

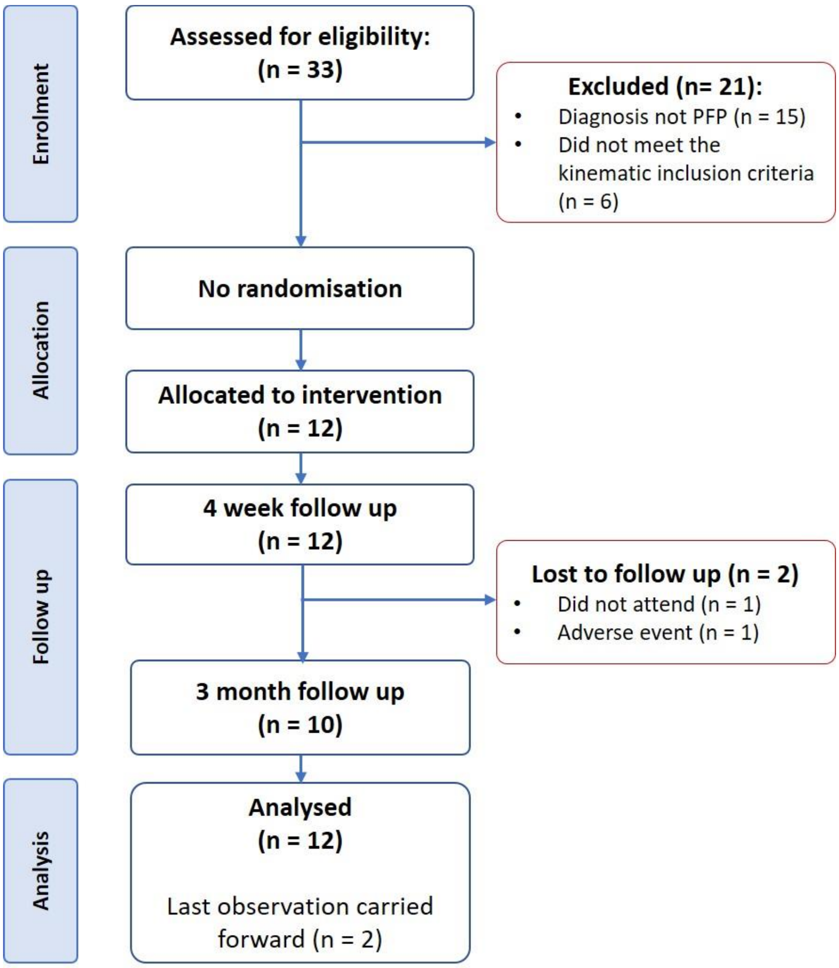
7.2.3 Kinematic data collection

Kinematic data were collected from all participants with confirmed PFP while running on a treadmill (Sole Fitness, F63, USA) at 3.2m/s wearing their own running shoes in accordance with procedures outlined in Section 3.3. After a 5-minute warm up period, 30 seconds of kinematic data were collected using a 12 camera Qualysis Oqus system (240Hz) (Gothenburg, Sweden). A total of nine anatomical segments were tracked following a previously published protocol (136, 201). Segments included the thorax, pelvis and bilateral thigh, shank and foot segments. Further details of the markers used to track each segment and the precise definition of the anatomical coordinate systems is provided in previous publications (136, 201, 235) and Chapter 3 Section 3.3.

Raw kinematic data were low pass filtered at 10Hz. Intersegmental kinematics, along with the motions of the pelvis and thorax with respect to the laboratory system, were calculated using a six degrees of freedom model using the Visual 3D (C -Motion, USA) software. Gait events were defined using a kinematic approach where initial contact was defined as the first vertical acceleration peak of either the heel or metatarsal markers and toe off defined as the vertical jerk peak of the 2nd metatarsal marker (363). Gait events were subsequently used to segment each kinematic signal into a minimum of 10 consecutive gait cycles. An ensemble average for each signal was created and selected kinematic parameters derived from the ensemble average curves. This latter processing was carried out using a custom Matlab script.

Participants were invited to participate in the gait retraining study providing they demonstrated hip and/ or pelvis kinematics in a range similar to that identified within Chapter 5, as being associated with running injuries. Specifically, inclusion criteria for hip and pelvis kinematics were defined as peak hip adduction (HADD) and/ or contralateral pelvic drop (CPD) angles equal to or greater than one standard deviation above the mean value of the control group reported in Chapter 5 (406) (qualifying criteria = CPD ≥5.6^{o}, HADD ≥13.2^{o}). This kinematic inclusion criteria was utilised based on findings of the literature review (Chapter 2, Section 2.4.7) suggesting clinical and biomechanical outcomes may be improved by specifically targeting interventions to those who demonstrate sub-optimal running kinematics at baseline. The kinematic parameters of CPD and HADD were selected based on their association with PFP reported in Chapter 5 (Section 5.3). Runners who did not meet the kinematic inclusion criteria were not included in the study and were referred to a health care professional for further management. Figure 27 provides an overview of participant inclusion and progression throughout the trial.

Figure 27: CONSORT flow diagram outlining participant progression throughout the trial.



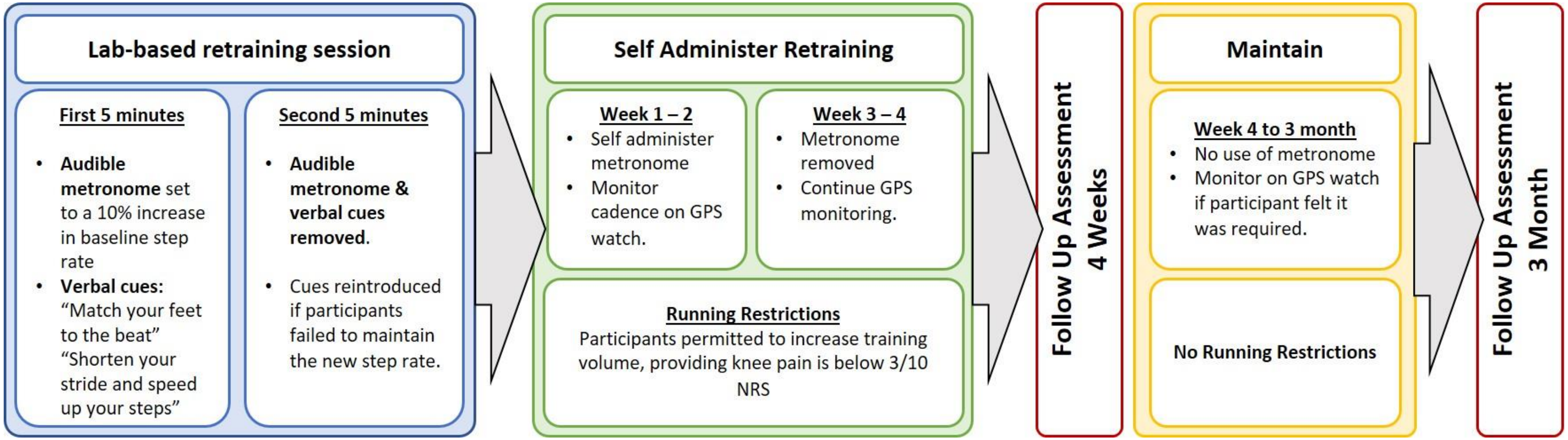
7.2.4 Retraining Protocol

All participants included within the retraining protocol completed a single 10-minute retraining session conducted immediately after the initial 3D gait analysis. A visual overview of gait retraining procedures is provided in Figure 28. During the retraining session participants were asked to run at the same speed with a 10% increase in their original step rate. Step rate was calculated as the number of foot contacts per minute. During the first 5 minutes of the retraining protocol, participants were instructed to match their footsteps to an audible metronome set to the new step rate. For the final 5 minutes of the retraining session, the audible metronome was removed and participants were instructed to continue running at the increased step rate. Throughout this time, participants were monitored by the lead researcher to ensure they were able to maintain the higher step rate, and the metronome reintroduced if they failed to do so.

Following the retraining session participants were provided with instructions for self-administration and monitoring of the increased step rate (Figure 28). Specifically, during the first two weeks, participants were instructed to continue using a freely downloadable metronome app set to the new step rate. During the third- and fourth-week participants were instructed to continue running without the use of the metronome but were instructed to self-monitor their cadence using their GPS watch. Participants were permitted to increase their running volume at any point in the retraining period providing any knee pain experienced was rated below 3/10 on an NRS scale.

All participants were invited to follow up 3D gait analysis sessions at 4 weeks and 3 months post initial assessment. This follow up period allowed us to investigate whether kinematic changes could be maintained across a time frame comparable to previous gait retraining studies (69). The follow up sessions were completed following the same kinematic testing procedures as visit one, recording the same clinical outcome measurements. Following the 4 week follow up assessment, participants were instructed to continue running without the use of the metronome. No restrictions to training parameters were provided, participants were instead instructed to increase their training volume, paces and change surfaces as they saw fit, providing any pain experienced was rated lower than 3/10 NRS.

Figure 28: Visual overview of gait retraining procedures and follow up timepoints.



7.2.5 Data Analysis

Several kinematic parameters were selected for data analysis. Kinematic parameters included peak contralateral pelvic drop, hip adduction, hip internal rotation and knee flexion angle. These parameters were selected based on their association to PFP reported in Chapter 5 and previous research highlighting associations with these parameters and PFP (58, 192, 404, 406). Peak angles at during stance were defined as the maximum joint angle between initial contact and toe off. Stride rate was also included within the analysis measured as steps per minute (spm), along with clinical outcome measures of worst pain experienced in the past week using the NRS, longest distance run pain free, total weekly running volume and LEFS score.

7.2.5.1 Statistical Analysis

One-way repeated measures ANOVA was used to assess for differences in kinematics parameters between initial assessment, 4 week follow up and 3 month follow up, with a critical alpha of <.05. When significant differences were observed, post hoc Bonferroni testing, adjusted critical alpha <.016, was used to identify differences between time-points. Clinical outcomes of NRS and LEFS were analysed using Friedman test for non-parametric data with post hoc Wilcoxon signed-rank test. Effect sizes were calculated for pairwise comparisons using Cohen's D and interpreted as 0.2, 0.5 and 0.8 as small, medium, and large, respectively (376).

Prior to the ANOVA, the assumption of sphericity was first assessed using Mauchly's test. Only peak knee flexion violated the assumption of sphericity ($\chi^2(2) = 9.2$, $p < .01$). When the assumption of sphericity is not met the F-statistic is positively biased increasing the chance of type 1 error (375, 407). Therefore, to account for the loss of sphericity, peak knee flexion was analysed using the Greenhouse-Geisser correction. The Greenhouse-Geisser correction was used instead of the Huynd-Feldt correction based on prior recommendations for its use when the estimate of sphericity produced is below .75 (Greenhouse-Geisser = .62) (375). Bonferroni post hoc testing was used as prior studies have reported this to be the most robust statistical method for repeated measurements with small sample sizes and when the assumption of sphericity is violated (375, 408).

Alternative methods such as Tukey's procedure have been reported to produce an inflated alpha level with multiple tests increasing the risk of type 2 error (375, 408).

7.3 Results

A total of 33 participants met the initial subjective inclusion criteria and were invited for a clinical examination (Figure 27). Following the clinical examination, a total of 18 were diagnosed as having PFP and invited to take part in the 3D gait analysis. Following the 3D gait analysis, a total of 12 participants met the inclusion criteria and were enrolled onto the gait retraining study. Two participants dropped out of the study between the 4 week and 3-month follow up points. The first subject failed to attend the 3-month follow up and did not respond to contact and the second developed a tibial stress fracture on the same limb and was unable to continue the study. Both participants were included in the final analysis using a last observation carried forward method (409) (Table 32).

Table 32: Participant characteristics. Values are presented as mean & standard deviation

Male/ Female	Age (years)	Body Mass (kg)	Height (cm)	Average weekly running volume (km)
4/8	39.9 (6.5)	61.0 (6.5)	170.3 (7.0)	29.0 (8.1)

7.3.1 Kinematics

One-way repeated measures ANOVA showed a significant effect of time for several kinematic parameters (Table 33). In particular, there were significant increases in step rate and reductions in peak CPD, HADD and knee flexion following the step rate intervention (Table 33, Figure 29 & Figure 30). Post hoc test revealed that step rate significantly increased by an average of 11.2% at 4 weeks (Mean Difference [MD], 18.6spm; 95% Confidence Interval [CI], 11.9spm, 25.2spm) and 9.2% at 3-month (MD, 15.1spm; 95% CI, 10.6spm, 19.6spm) when compared to baseline. There was a significant 3.1^o reduction in CPD (MD, 3.1^o; 95% CI, 1.9^o, 4.4^o) (Figure 29) and 3.9^o reduction in HADD (MD, 3.9^o; 95% CI, 2.0^o, 6.0^o) (Figure 30) at 4 week follow up, which was also significant at 3-month when compared to baseline for both CPD (MD, 2.7^{o}; 95% CI, 1.4^{o}, 4.1^{o}) and HADD (MD, 2.8^{o}; 95% CI, 0.4^{o}, 5.4^{o}). Similarly, there was a significant 4.1^{o} reduction in peak stance phase knee flexion at 4 weeks (MD, 4.1^{o}; 95% CI, 0.1^{o}, 8.2^{o}) and 3-month (MD, 4.1^{o}; 95% CI, 0.8^{o}, 7.5^{o}). No significant differences were observed between the 4 week and 3-month follow up time points for any of the kinematic parameters (Table 33).

*Table 33: Mean [SD] kinematic parameters at initial assessment, 4 week and 3 month follow up. * indicates significant difference when compared to baseline at p <.016 following Bonferroni correction. HADD = peak hip adduction, HIR = peak hip internal rotation, CPD = peak contralateral pelvic drop, KFlx = peak knee flexion. Pairwise effect sizes are presented using Cohen's D and interpreted as 0.2 = small, 0.5 = medium, 0.8 = large, IA = initial assessment.*

	IA	4 Week	3 Month	P value	Pairwise Effect sizes (Cohen's D)	
Stride Rate (Steps per minute) (spm)*	165.9 (7.4)	184.5 (10.1)*	181.0 (7.8)*	<.01*	**IA to 4 Week:**	2.1
					IA to 3 month:	1.9
					4 week to 3 month:	0.4
CPD $(^{o})$*	7.5 (1.8)	4.3 (2.5)*	4.7 (3.0)*	<.01*	**IA to 4 Week:**	1.4
					IA to 3 month:	1.1
					4 week to 3 month:	0.1
HADD $(^{o})$*	15.9 (2.8)	11.9 (1.8)*	13.1 (3.2)*	<.01*	**IA to 4 Week:**	1.7
					IA to 3 month:	0.9
					4 week to 3 month:	0.4
HIR $(^{o})$	4.1 (8.2)	4.2 (9.4)	4.4 (7.9)	.93	**IA to 4 Week:**	0.0
					IA to 3 month:	0.0
					4 week to 3 month:	0.0
KFlx $(^{o})$*	33.7 (5.3)	29.6 (3.2)*	29.5 (3.2)*	<.01*	**IA to 4 Week:**	0.9
					IA to 3 month:	0.9
					4 week to 3 month:	0.0

*Figure 29: Ensemble group average curve for frontal plane pelvis kinematics across the stance phase. Solid line represents group mean, shaded area represents 1SD. X-axis = percentage of stance phase. Y-axis = frontal plane pelvis angle in degrees, +ve values indicate contralateral pelvis drop, -ve values indicate contralateral pelvis elevation. *indicates statistically significant between groups.*

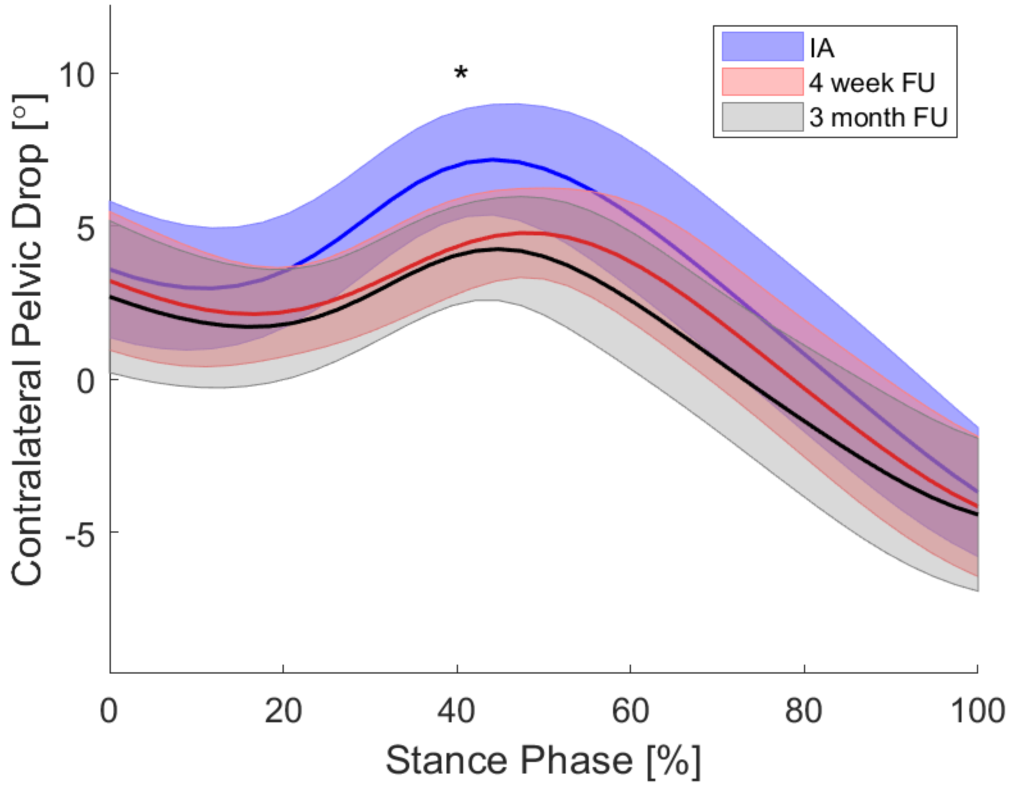
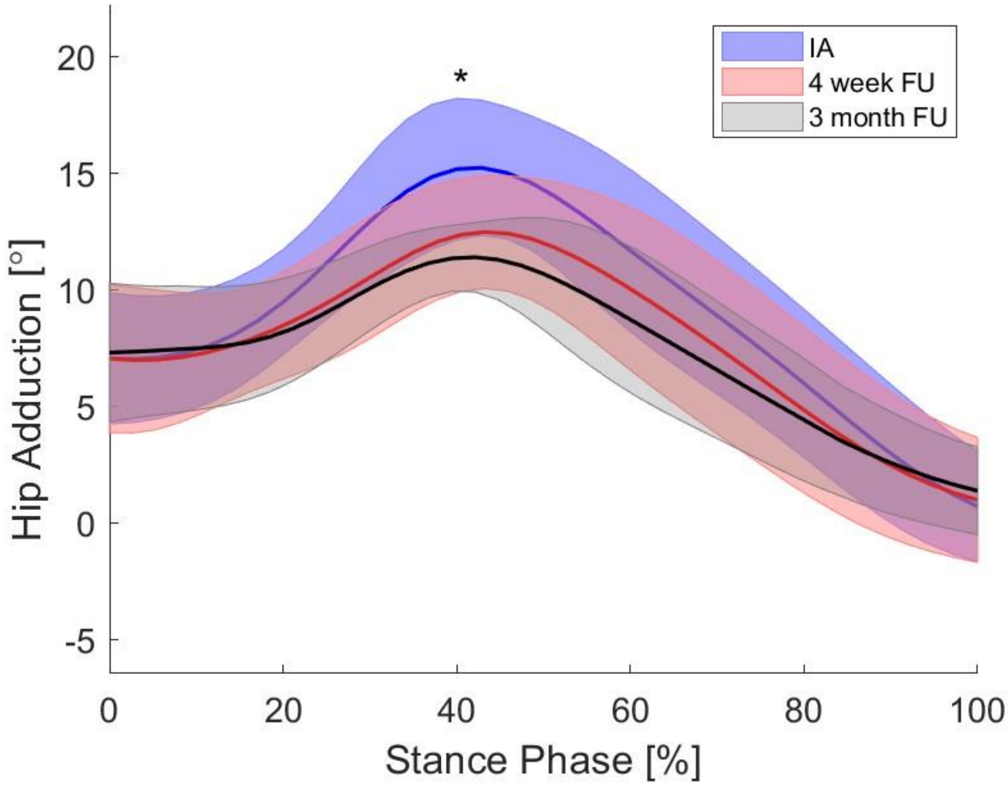


*Figure 30: Ensemble group average curve for frontal plane hip kinematics across the stance phase. Solid line represents group mean, shaded area represents 1SD. X-axis = percentage of stance phase. Y-axis = frontal plane hip angle in degrees, +ve values indicate hip adduction, -ve values indicate hip abduction. *indicates statistically significant between groups.*



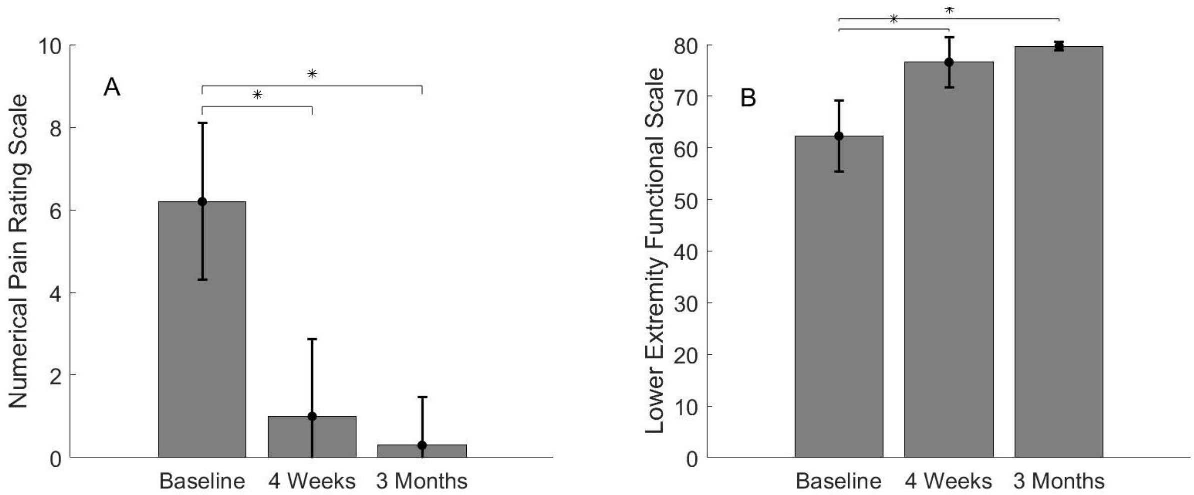
7.3.2 Clinical and Functional Outcomes

All clinical and functional outcomes demonstrated statistically significant improvements. Specifically, there was a significant reduction in pain scores on the NRS from an average of 6.2/10 at baseline to 1.0 and 0.3 at 4 weeks and 3-months respectively (χ^2 = 21.38, p <.01) (Figure 31) which is above the minimal clinically important difference of 1.2 points (410). LEFS demonstrated a statistically significant improvement from 62.3 at baseline to 76.6 at 4 weeks and 79.7 at 3-months (χ^2 = 22.29, p <.01) (Figure 31). When compared to baseline this was a 14.3 point and 17.4 point improvement at 4 weeks and 3-months respectively, which is above the minimal clinically important difference of 9 points (326). All participants demonstrated a significant increase in total weekly running volume (MD, 13.8km; 95% CI, 4.6km, 22.9km) and longest distance run pain free (MD, 6.8km; 95% CI, 3.1km, 10.6km) from baseline values to 4 week and 3-month follow up (Table 34).

*Table 34: Mean [SD] Functional outcome measures at initial assessment, 4 week and 3 month follow up. * indicated statistically significant difference when compared to baseline at p <.016. Pairwise effect sizes are presented using Cohen's D and interpreted as 0.2 = small, 0.5 = medium, 0.8 = large, IA = initial assessment.*

	IA	4 Week	3 Month	P value	Pairwise Effect size (Cohen's D)	
Total distance per week (km)*	13.3 (9.8)	27.1* (11.3)	28.3* (13.0)	<.01	**IA to 4 Week:**	1.3
					IA to 3 month:	1.3
					4 week to 3 month:	0.1
Longest Run Pain Free (km)*	2.0 (1.2)	8.9* (4.4)	11.3* (6.4)	<.01	**IA to 4 Week:**	2.1
					IA to 3 month:	2.0
					4 week to 3 month:	0.4

*Figure 31: Clinical outcome measures at initial assessment, 4 week and 3 month follow up. * indicated statistically significant difference when compared to baseline at p <.016. Error bars represent +/- 1 standard deviation.*



7.4 Discussion

The objective of this study was to investigate whether a 10% increase in running step rate influences frontal plane kinematics of the hip and pelvis, as well as clinical outcomes in runners with PFP. In support of our hypothesis, we observed significant reductions in frontal plane pelvis and hip kinematics, as well as significant reductions in pain, improvements in function and running at 4 weeks, which appeared to be maintained at 3-month follow up.

7.4.1 Kinematic Response to Retraining

Following the step rate increase, we observed a 3.1^{o} and 3.9^{o} reduction in CPD and hip adduction (Table 33, Figure 29 & Figure 30), which may offer a mechanical explanation for the improved clinical outcomes seen in this study. These changes are greater than that observed in previous step rate studies (255, 267), with this the first study to highlight kinematic adaptations are maintained at longer term follow up. In Chapter 5, we identified both of these parameters to be associated with PFP, with CPD identified as a kinematic parameter consistently observed across multiple different injuries. This is in agreement with prior studies citing contralateral pelvic drop and hip adduction as kinematic risk factors for PFP (57, 58, 185, 192, 406). These kinematic patterns are thought to increase tissue loads per stride of a run, which when combined with an exposure to external training loads, may result in a cumulative tissue stress which exceeds tissue capacity leading to injury development.

In relation to PFP, peak CPD and HADD may contribute to elevated tissue loads at the patellofemoral joint via several mechanisms. It is thought that contralateral pelvic drop will give rise to an increase in iliotibial band tension resulting in lateral displacement of the patella (178, 180), while hip adduction would cause the femur to shift medially under the patella (174). This would result in elevated contact pressure between the patella and the lateral facet leading to elevated joint stress and potential injury development (168). Therefore, it is possible that the observed reductions in CPD and HADD following an increase in step rate, could contribute to reduced lateral displacement of the patella and a corresponding reduction in patellofemoral joint stress.

Similarly, the reduction in peak stance phase knee flexion may also contribute to improvements in clinical outcomes. Peak stance phase knee flexion has been shown to influence patellofemoral joint reaction force, explaining up to 64% of the variance in peak patellofemoral joint load (51). Smaller knee flexion angles at mid stance will likely reduce the external joint forces as well as reduce the demand on the surrounding musculature (51). In the current study we observed a 4.1^{o} reduction in peak knee flexion (Table 33). Given the work of Lenhart et al (51) this magnitude of change is likely to contribute to reductions in peak patellofemoral joint force. These reductions, combined with the reductions in peak hip adduction and CPD, will likely lead to significant reductions in patellofemoral joint stress which may explain the observed improvements in clinical outcomes within the present study. However, it is important to note that not all participants demonstrated a reduction in peak knee flexion (Table 35, Figure 32). Out of 12 participants only 4 demonstrated a reduction in peak knee flexion angle greater than the minimal detectable change (MDC) of 5^{o} reported in Chapter 4 (Table 35). Conversely, 9 participants demonstrated reductions in CPD and 10 for peak HADD which exceeded the MDCs of 1.7^{o} and 1.8^{o} (Table 35). Therefore, it is possible that changes to peak knee flexion angle are unlikely to explain the clinical improvements observed across all participants.

We suggest that the improved frontal plane hip and pelvis kinematics may be explained by alterations in neuromuscular activity of the hip. Willson et al (138) found that runners with PFP demonstrate significantly delayed onset of the gluteus medius, which had a moderate correlation with hip adduction excursion. It is hypothesised that delayed muscle activation of the gluteus medius during the stance phase of running would result in a loss of neuromuscular stiffness about the hip and pelvis leading to a loss of frontal plane stability (138). Increasing step rate by 10% has been shown to directly influence the preactivation of the gluteal muscles (411). Specifically, Chumanov et al (411) reported significantly increased gluteus medius and maximus muscle activity in late swing, just prior to initial foot contact following a 10% increase in step rate. Considering the role gluteus medius plays in frontal plane stability of the hip and pelvis, it is likely that the earlier onset of the gluteal muscles would result in increased neuromuscular stability during the stance phase of gait. This would likely explain the mechanical improvements of reduced CPD and hip adduction observed in the present study.

Reductions in peak knee flexion may also be explained by alterations in neuromuscular activity at the knee. Increasing step rate has been shown to result in greater preactivation of the hamstrings, vastus lateralis and rectus femoris during late swing (411). It is thought that these changes in neuromuscular coordination contribute to a more extended knee throughout the stance phase, reducing peak knee flexion angles (51, 411).

In contrast to previous studies we did not identify differences in peak hip internal rotation following gait retraining. Neal et al (255) reported a 5.1^{o} reduction in peak hip internal rotation following a 10% increase in step rate, whereas in the present study we did not observe more than a 0.5^{o} change. This may be explained by our baseline inclusion criteria of increased hip adduction and/ or contralateral pelvic drop, rather than hip internal rotation. Participants within this study demonstrated 4.1^{o} of hip internal rotation at baseline, which is less than the 9.1^{o} reported in the study by Neal et al (255) and similar to the 4.4^{o} reported amongst the healthy runners in Chapter 5. Therefore, it is possible that participants in the present study did not demonstrate increased hip internal rotation angles at baseline and thus would be unlikely to demonstrate any change.

7.4.2 Magnitude of Change

An interesting finding was the magnitude of clinical improvements made by participants. Specifically, participants reported their worst pain to be on average 1.0 out of 10 at 4 week follow up and 0.3 out of 10 at 3 month follow up (Figure 31). This is greater than the minimum clinically important difference of 1.2 points (410) and greater than improvements seen in previous step rate studies, which have reported average NRS scores of 3.9 (267), 3.8 (318) and 2.9 out of 10 (255) post retraining. We also observed significant improvements in function with all runners reporting an increase in their weekly running volume and longest distance run pain free as early as 4 weeks (Table 34), as well as a 17.4 point LEFS improvement at 3 months, exceeding the minimum clinically important difference of 9 points (326). This contrasts to previous step rate studies, with one study reporting participants to be running less than their pre-injury status at 20-week follow up (318) and another study reporting less than a 9 point improvement on the LEFS (267).

The reason for the magnitude of kinematic and clinical improvements in the present study compared to previous step rate studies may be due to the differences in inclusion criteria. In the present study we specifically targeted participants who demonstrated sub-optimal kinematics at baseline. We did this to account for the multifactorial aetiology of PFP and ensure the appropriate underlying injury driver was targeted through the gait intervention. Failure to consider alternative causes of injury would likely result in the inclusion of biomechanical non-responders within the retraining group. As such, these participants would be unlikely to demonstrate significant clinical improvements. Willy et al (69) and Noehren et al (70) are the only previous studies to use a similar inclusion criterion, with their results showing a similar magnitude of clinical improvement. Therefore, we would suggest that future research should aim to establish the underlying pathological driver in order to appropriately target clinical interventions, and that this be mirrored in clinical practice.

A secondary objective of this study was to investigate whether runners can self-administer a 10% increase in step rate using an audible metronome and a GPS smart watch, and whether these changes can be maintained at short term and long term follow up. The results do support this hypothesis as on average, runners demonstrated an increase in step rate and improvement in hip and/or pelvis kinematics at 4 week follow up, which was maintained at 3 month follow up. At 4 week follow up there was a 3.1^{o} reduction in CPD and 3.9^{o} reduction in HADD which exceeded the minimal detectable change values of 1.7^{o} and 1.8^{o} reported in Chapter 4 (Table 23) and therefore, likely represents true intervention effects. Although there was a subtle change in peak CPD (0.4^{o}) and HADD (1.2^{o}) towards baseline values at 3 month follow up, the magnitude of the of the differences was typically small and did not exceed the MDC reported in Chapter 4 (Table 23). Therefore, we feel subtle differences between these time points are unlikely to be clinically important. However, we acknowledge that the small sample size limited the statistical power to detect small differences between these time points. Future randomised control trials with larger participant numbers are now necessary to further validate our findings and confirm that kinematic changes are maintained over longer time periods.

7.4.3 Clinical Relevance

In contrast to previous gait retraining studies we opted to allow runners to self-administer and self-monitor their retraining using a metronome app and feedback from a GPS smart watch. This proved successful as all runners were able to maintain an increased cadence at 4 week and 3-month follow up. Furthermore, at 4 week follow up all participants reported they did not use the metronome beyond the first week and instead would self-monitor their cadence using their GPS watch. Previous studies have utilised a faded feedback design where feedback is gradually removed over 8 sessions across a 2 week period. Although faded feedback designs have proven clinically effective, they require close clinical supervision and are restricted to clinical and laboratory settings. The present study demonstrates that simple step rate retraining can be applied outside of the laboratory and with minimal clinical contact. Importantly, two dimensional measures of contralateral pelvic drop and hip adduction have been shown to be valid and reliable when compared to three dimensional measurements (389). Therefore, assessment of running kinematics and gait retraining could be integrated into clinical practice and a participant's normal running routine.

7.4.4 Individual Step Rate Variability

Interestingly, upon inspection of individual data there appears to be considerable variation between participants in their ability to attain the desired step rate (Table 35, Figure 32, Figure 33 & Figure 34). Although the group average increase was 11.2%, the variability between participants ranged between 5.1% and 18.1%, with only 2 participants attaining the target 10% increase. This could be explained by the limited external feedback provided to participants through only a single retraining session. However, these findings are similar to that of Neal et al (255) who utilised a faded feedback design to increase step rate by 7.5% yet reported the attained step rate to range between 2.3% and 11.1%. Therefore, the results from both studies highlight considerable variability in the ability of participants to attain the target step rate. It is possible that some runners may be more amenable to motor learning than others, consequently for some individuals it may be necessary to provide greater feedback during the retraining process in order to facilitate the acquisition of the desired step rate change.

Despite the variability in the percentage increase in step rate, this did not appear to influence the magnitude of change in kinematics (Figure 32, Figure 33 & Figure 34). For example, despite only a 6.3% increasing in step rate, participant 5 (Table 35) demonstrated the greatest reduction in peak hip adduction of all participants with a 7.6^{o} reduction when compared to baseline (depicted in light green in Figure 34). Therefore, it is possible that achieving a 10% increase in step rate is not necessary for improvements in kinematic patterns. Instead the magnitude of kinematic change may be influenced by additional factors such as the baseline kinematic values or individual changes in muscle activation patterns (411). Future studies should therefore consider further investigating underpinning reasons for the magnitude of kinematic changes in response to step rate retraining.

Table 35: Individual participant response to step rate retraining at 4 week follow up assessment. CPD = contralateral pelvic drop, HADD = hip adduction, KFlx = knee flexion.

Participant	Baseline step rate	Change in Stride Rate (%)	Change in CPD (º)	Change in HADD (º)	KFlx (º)
1	159	↑ 5.7	↓ 3.3	↓ 4.7	↓ 0.3
2	161	↑ 18.5	↓ 1.0	↓ 4.2	↓ 5.4
3	166	↑ 10.9	↓ 2.1	↓ 2.7	↑ 4.3
4	172	↑ 14.2	↓ 3.8	↓ 4.5	↓ 3.0
5	171	↑ 6.3	↓ 2.6	↓ 7.6	↓ 4.0
6	158	↑ 10.1	↓ 1.4	↓ 2.7	↓ 1.5
7	166	↑ 15.1	↓ 3.0	↓ 3.2	↓ 13.5
8	161	↑ 7.9	↓ 3.9	↓ 7.1	↓ 11.6
9	154	↑ 18.1	↓ 6.2	↓ 6.5	↓ 8.1
10	173	↑ 16.8	↓ 4.6	↓ 4.4	↓ 3.0
11	171	↑ 5.4	↓ 4.2	↑ 0.7	↓ 0.0
12	178	↑ 6.4	↓ 1.3	↓ 0.9	↓ 3.0

Figure 32: Individual data plots for the observed change in peak Knee Flexion (°) and step rate at 4 week follow up. Small circles represent values at initial assessment, larger circles represent values at 4 week follow up. Greater the vertical slope of the line between circles represents the magnitude of change in knee flexion angle. Horizontal distance of the line represents magnitude of change in step rate. Colours represent the individual participant data and are consistent throughout Figures 32, 33 and 24.

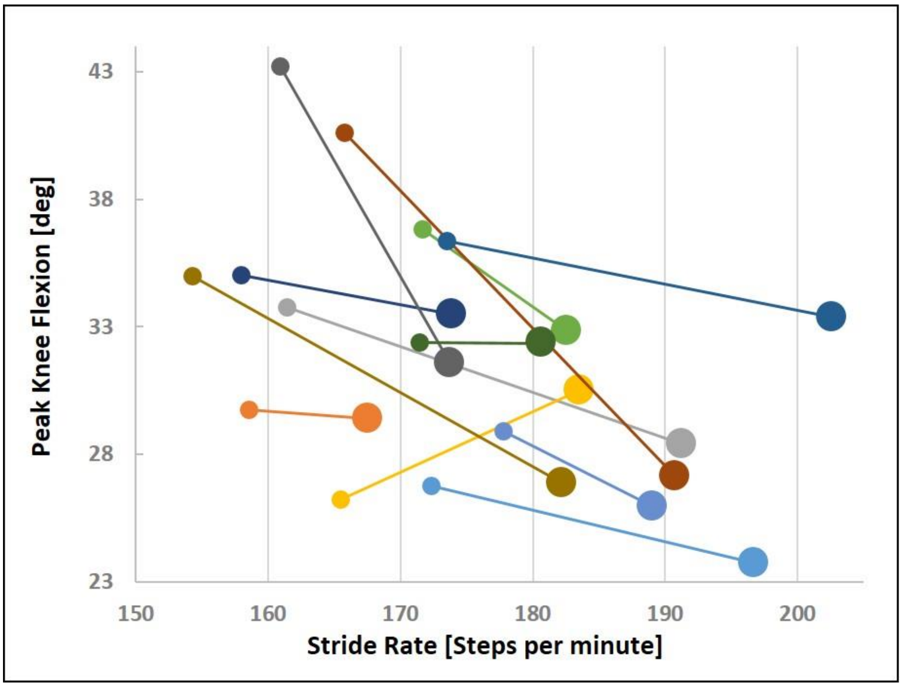


Figure 33: Individual data plots for the observed change in peak Contralateral Pelvic Drop (°) and step rate at 4 week follow up. Small circles represent values at initial assessment, larger circles represent values at 4 week follow up. Greater the vertical slope of the line between circles represents the magnitude of change in contralateral pelvic drop angle. Horizontal distance of the line represents magnitude of change in step rate. Colours represent the individual participant data and are consistent throughout Figures 32, 33 and 24.

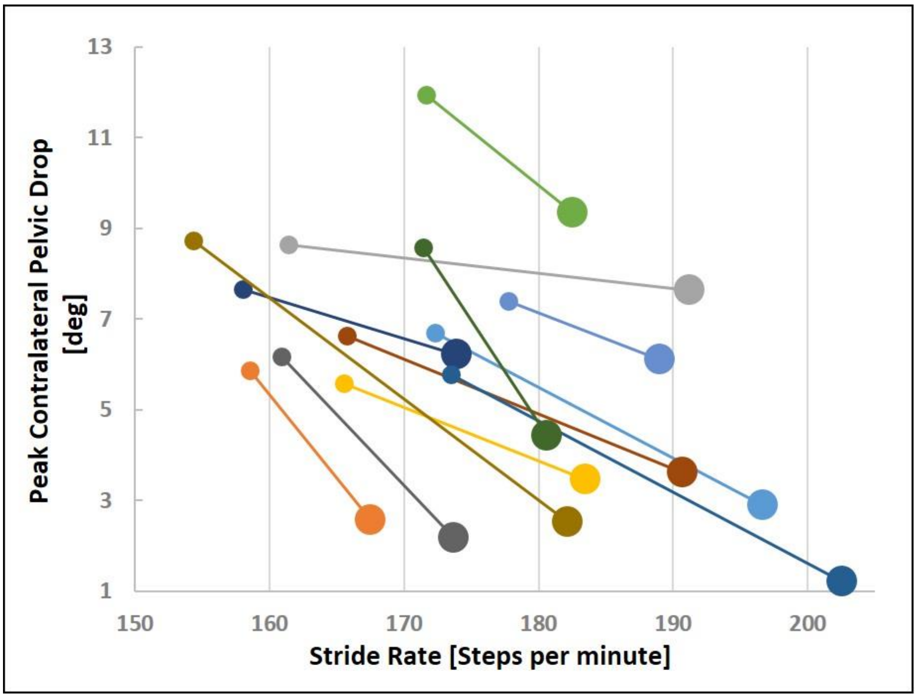
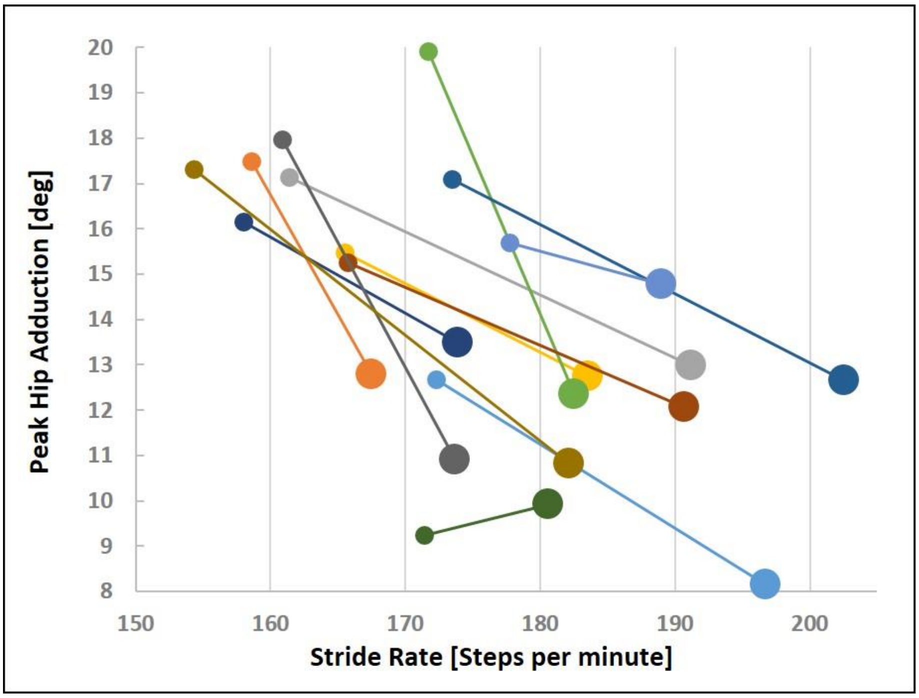


Figure 34: Individual data plots for the observed change in peak Hip Adduction (°) and step rate at 4 week follow up. Small circles represent values at initial assessment, larger circles represent values at 4 week follow up. Greater the vertical slope of the line between circles represents the magnitude of change in hip adduction angle. Horizontal distance of the line represents magnitude of change in step rate. Colours represent the individual participant data and are consistent throughout Figures 32, 33 and 24.

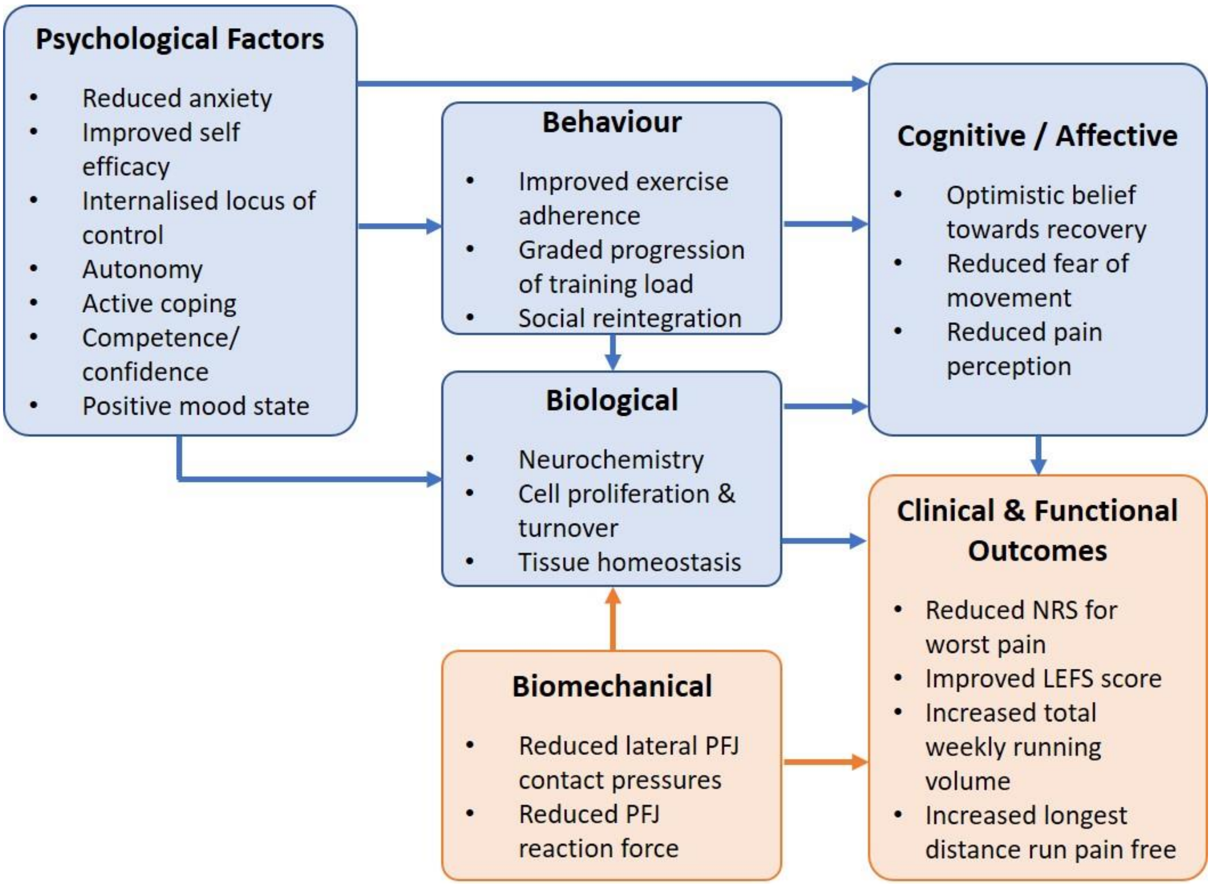


7.4.5 Limitations

One limitation of the present study is the lack of a control group making it difficult to ascertain whether the observed improvements in clinical symptoms are true intervention effects. Without a control group it is possible that symptoms may improve for reasons unrelated to the intervention such as regression to the mean or the natural recovery process of an injury (412). However, in order to account for this we recruited participants who had experienced symptoms for greater than 3 months in duration, which has been reported to be predictive of poor prognosis at long term follow up (413) (414). Therefore, we feel it is unlikely that participants would have experienced the magnitude of symptom improvement without clinical intervention. However, we acknowledge that without further randomised control trials we cannot rule out the possibility of confirmation bias within the interpretation of the present findings.

A second limitation is the possibility that the step rate intervention may have influenced rehabilitation outcomes via mechanisms not accounted for within this study. In the present study we focused upon biomechanical factors influencing recovery. However, rehabilitation outcomes are also known to be influenced by a variety of psychological, social and behavioural factors (Figure 35) (78, 79). Psychological factors such as an external locus of control, fear avoidance of pain associated with loading activities and negative perceptions towards ongoing pain have been linked to negative clinical outcomes and the persistence of pain (78, 79, 415, 416). It is possible that the step rate intervention provided, may have positively impacted a number of these factors (Figure 35). For example, through instructing participants to self-administer retraining sessions and self-progress their training volume, this may have facilitated positive psychological outcomes, such as an internal locus of control, active coping and improved self-efficacy. This may have, in turn, led to positive behavioural responses such as adherence to exercise and graded progression of training loads and reintegration with social peer groups, which may subsequently lead to positive biological and cognitive processes influencing the clinical and functional outcomes observed. Therefore, although the step rate intervention appears clinically effective, the intervention effects may extend beyond those that are biomechanical which should be acknowledged in clinical practise.

Figure 35: Conceptual diagram of the potential biopsychosocial influences to rehabilitation outcomes adapted from Brewer (78).
Orange boxes represent factors considered within the present study. Blue boxes represent additional biopsychosocial factors which may have influenced the rehabilitation outcomes observed. Arrows between boxes represent the interaction effects between multiple different factors.



As the lead researcher collected all patient reported outcome measures, it is possible this may have indirectly influenced participant reporting on subjective questionnaires and pain scales (412). The direct interaction between participant and researcher may have inadvertently led to the subjective reporting, or inflation, of treatment success amongst participants. This could have been influenced by positive interactions and a subconscious "desire to please" or conform to the perceived expectations of the researcher (412). Therefore, future intervention studies should consider utilising online feedback methods conducted after all testing procedures in order to limit potential bias introduced through the interaction between participants and the researcher.

Finally, it is important to note that one participant dropped out of the intervention after suffering a tibial stress fracture. This participant reported that injury onset occurred following a sudden increase in training volume in preparation for a half marathon. As we did not control participant progression of training volume, it is possible that the injury could be the result of training behaviours rather than a response to the intervention. As such, we would recommend that future clinical interventions provide participants with specific advice on the safe progression of running volume in order to reduce the risk of further injury.

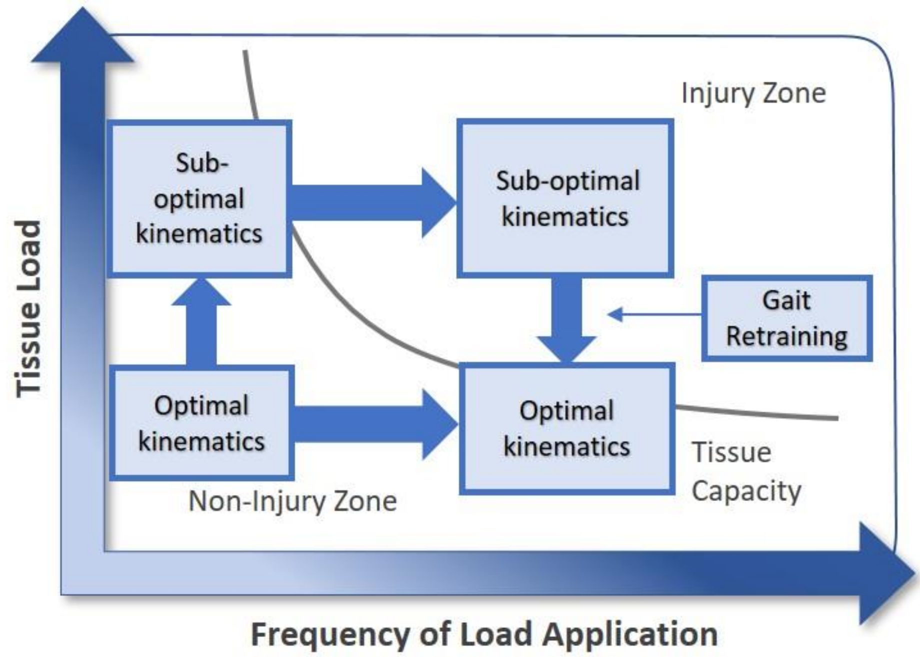
7.5 Summary and Implications

The results of this study highlight that a 10% increase in step rate improves running kinematics, clinical and functional outcomes at 4 weeks, which are maintained at 3 months amongst runners with PFP. Targeting kinematic parameters associated with running related injuries may reduce tissue loads per stride, reducing the cumulative tissue loads applied to injured structures and appears to help improve clinical and functional outcomes amongst runners. Therefore, amongst injured runners step rate retraining appears to be a clinically effective intervention in the rehabilitation of PFP and can easily be integrated into clinical practice. Considering similar kinematics were observed across multiple different running injuries (Chapter 5), future studies should consider investigating the effectiveness of gait retraining interventions targeted to additional running related injuries.

8 Chapter 8: Discussion

Running biomechanics are often cited as a risk factor for running related injuries. It is thought that certain biomechanical parameters may increase the load applied to the musculoskeletal system during each stride of a run. When combined with an exposure to external training load, this will ultimately influence the cumulative tissue load encountered across a run or training period. If the cumulative tissue load exceeds tissue capacity this may result in injury development. In such instances, interventions which modify running biomechanics, could reduce tissue load per foot contact and therefore the cumulative tissue load across a run, assisting in the rehabilitation of injured runners. This process was outlined within the introduction Section using Figure 2 as an illustrative example (reprinted here as Figure 36) which motivated the overarching aim of this thesis. This aim was to identify biomechanical characteristics associated with common running injuries, explore whether training load exposure influences running kinematics and, finally, investigate whether gait retraining can be used to improve biomechanics, clinical and functional outcomes amongst injured runners. In order to achieve this aim, this thesis first conducted a literature review to identify kinematic and kinematic parameters associated with common running injuries.

Figure 36: Adapted stress frequency curve from Hreljac & Ferber (1). The Figure is used as a conceptual diagram illustrating the influence of kinematics upon tissue load, how the interaction with external training loads may influence injury development and the role of gait retraining.



8.1 Biomechanical characteristics associated with common running injuries

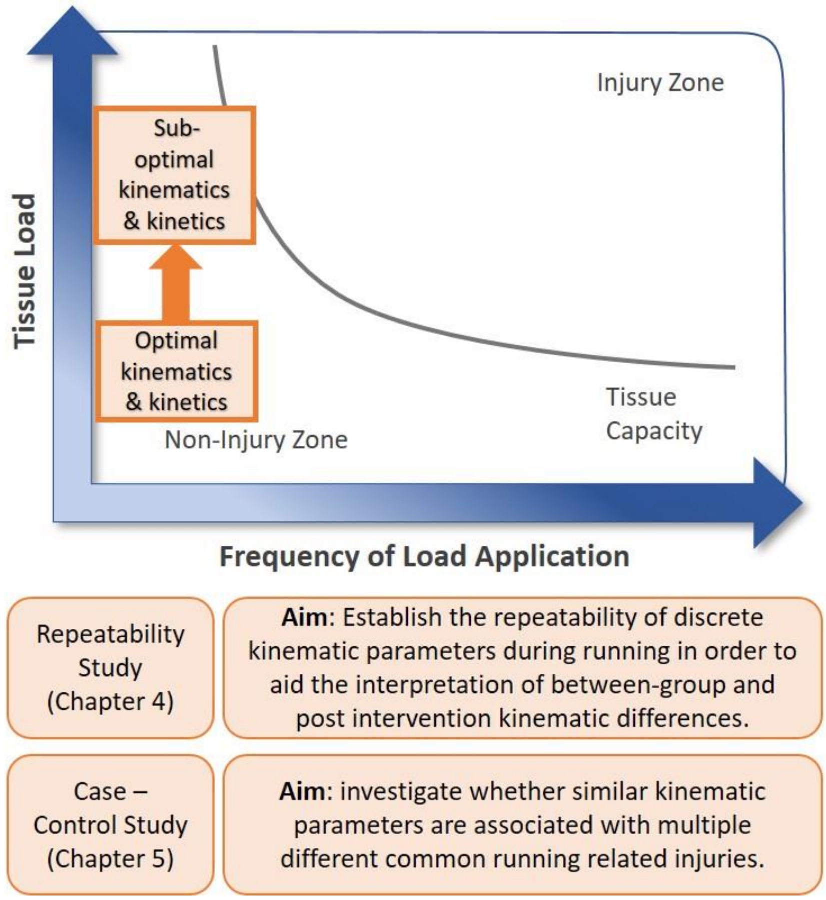
Within Chapter 2, the literature review identified limited evidence to support an association between kinetic parameters and common running related injuries (Table 17). Of the available evidence it appears that only tibial stress fractures demonstrate a consistent association with the kinetic parameter of vertical loading rate, with a lack of evidence identified for an association between kinetic patterns and the common running injuries of PFP, ITBS and AT (Table 17). Although two prospective studies have reported greater horizontal breaking forces and vertical loading rates to be linked to future injury development (53, 60), the number of pathology specific cases identified within these studies are relatively small, with limited additional studies supporting an association between kinetic patterns and running injuries of PFP, AT and ITBS. This is an interesting finding considering a large body of research has focused on gait interventions specifically targeted toward reducing of vertical loading rate through transitioning to a forefoot strike running pattern, or using real time feedback of impact forces (399, 417-

419). While such interventions may be beneficial for runners at risk of, or with a history of tibial stress fractures, findings from the current literature review suggest such methods may not target mechanical patterns underlying other common running injuries, such AT, ITBS and PFP.

In contrast to kinetic patterns, several similar kinematic parameters were found to be associated with multiple different running related injuries (Table 18). For example, peak hip adduction and internal rotation were found to be associated with MTSS, PFP and ITBS in several prospective and retrospective studies. Suggesting that there may be similar kinematic parameters that could increase the load applied to multiple different musculoskeletal structures during running and could represent global contributors to common running related injuries. Identification of such parameters could be of clinical value as they could ultimately be targeted through gait interventions. Subsequently such parameters could be targeted within rehabilitation and injury prevention programs, benefiting multiple different running injuries rather than being limited to single pathologies.

These findings motivated the initial objectives set out in Chapters 4 and 5. First, in order to aid the interpretation of between-group and post intervention kinematic differences, it was deemed necessary to establish the repeatability of testing procedures. Therefore, the objective of Chapter 4 was to investigate the between day repeatability, standard error of measurement and minimal detectable change of discrete kinematic parameters of the trunk, pelvis and lower limbs during treadmill running. After investigating this repeatability, the objective of Chapter 5 was to investigate whether similar kinematic parameters are associated with multiple different common running related injuries. The impact of achieving these two objectives was to meet the first part of the overarching aim, specifically to identify biomechanical characteristics associated with common running injuries (Figure 37).

Figure 37: Using the adapted stress frequency curve from Hreljac & Ferber (1), the Figure provides an illustration of the knowledge gaps which relevant chapters of the thesis aimed to address. Orange boxes represent the aims of the specific chapters and the specific knowledge gap addressed through fulfilling these aims.



8.2 The repeatability of kinematic measurements.

In agreement with several previous studies (199, 253), the repeatability study found peak transverse plane joint angles of the hip and knee, as well as peak rearfoot eversion to demonstrate the greatest measurement error of all kinematic parameters. The large measurement errors are likely to have implications for the interpretation of between group differences and the effect of clinical interventions. Current theoretical concepts suggest that rearfoot eversion and hip internal rotation may play a role in common injuries such as AT, PFP, ITBS and MTSS (50, 94, 118, 145). However, the literature review identified limited and conflicting evidence to support this association (Table 18). One possibility is that the measurement errors associated with these parameters could explain the conflicting findings in the literature.

Previous authors have argued that, for biomechanical measurement systems to be useful in clinical practise, they must be able to produce stable, repeatable results and accurately represent the movement of interest (350, 420). The results from the present work and that of previous studies would suggest that measurements of hip internal rotation and rearfoot eversion do not meet these criteria. In Chapter 4, peak hip internal rotation and rearfoot eversion were found to demonstrate SEMs of 3.2^o and 2.3^o which represents approximately 54% and 71% of the total peak values. Measurement errors of such magnitude suggest a high level of noise is present within these data. Consequently, this may limit the ability to evaluate the effect of clinical interventions and reduce statistical power to detect small, potentially meaningful between-group differences. Further studies have also questioned the validity of transverse plane measurements of the hip using skin mounted markers and rearfoot kinematics using shoe mounted markers, suggesting these measurement techniques do not accurately represent the underlying skeletal movement (131, 202). Therefore, even if theoretical concepts and modelling studies appear to provide a plausible link between these parameters and mechanisms of tissue stress, the limited repeatability, large measurement errors and questionable validity associated with these measurements, may limit the ability to accurately identify associations between the two.

In contrast to transverse plane kinematics, frontal and sagittal plane kinematics were found to demonstrate good to excellent repeatability with low SEMs and MDCs. Interestingly the SEMs and MDCs are similar to studies investigating the repeatability of discrete kinematic parameters utilising 2D measurements. Using 2D measurements, Dingenen et al (392) reported an SEM of 1^o and MDCs of 2.7^o and 2.8^o for peak CPD and HADD, whereas using 3D measurements in Chapter 4, the SEM for peak CPD and hip adduction was 0.6^o and 0.7^o with MDCs of 1.7^o and 1.8^o respectively. This offers potential clinical implications, as the comparable reliability of 2D and 3D systems means the 2D assessment of running kinematics could be integrated into clinical practice without the need for expensive 3D measurement systems.

Although utilising 2D measures offers a practical method for assessment of running kinematics, there are several barriers limiting the wide-spread clinical application of assessment methods. Firstly, limited evidence has investigated the validity and accuracy of 2D measurements when compared to 3D measurements. Second, whether comparable results can be obtained from multiple testing sites is also uncertain (259). In order to make clinical decisions regarding what constitutes sub-optimal or optimal kinematics, it is necessary to compare individual participant data to a reference “normative” dataset. Therefore, if 2D measurements demonstrate large between site variation in results and lack validity when compared to 3D measurements, then practical application of testing could lead to inaccurate and misleading conclusions. Consequently, further studies are now required to compare the validity of 2D measurements of kinematic measurements to those obtained from 3D systems and whether repeatable measures can be obtained across multiple testing locations. In doing so, this could allow for clinical integration of 2D assessment measures using “normative” reference datasets published from large scale studies.

8.3 Kinematic characteristics associated with common running injuries

Having established the repeatability of kinematic testing procedures, Chapter 5 sought to investigate whether similar kinematics are associated with multiple different running related injuries. Accepting the experimental hypothesis, Chapter 5 found peak contralateral pelvic drop (CPD) and forward trunk lean, as well as a more extended knee and dorsiflexed ankle at initial contact to be associated with multiple common running injuries. Whereas previous studies have focused upon kinematic patterns associated with specific injuries, the present study highlights that certain kinematic parameters may increase tissue loads placed upon multiple different musculoskeletal structures.

Interestingly, peak CPD was identified to be the kinematic parameter most strongly associated with common running injuries. This parameter appeared to be consistent across all injury subgroups and did not appear to be influenced by gender. This is in contrast to other parameters such as peak hip adduction, which although associated with multiple different running injuries (Chapter 2, Table 18), has been suggested to be more strongly associated with the female sex (58, 194, 238). Considering the consistency of peak CPD across both genders and across the injury subgroups, this parameter may represent a global kinematic contributor to running related injuries. The novel contribution of this finding is that this information could be used to inform injury prevention and rehabilitation programs targeted to a range of common running injuries across both genders, rather than being limited to specific subgroups.

An interesting question raised by these findings is why certain tissues become injured rather than others. This perhaps reflects the multifactorial nature of running related injuries. Although kinematics may increase tissue loads throughout the musculoskeletal system, whether injury occurs, and the specific tissues which become injured, is likely to depend upon whether the tissue load encountered exceeds the capacity for each tissue to tolerate load. As outlined within the introduction, tissue-specific load capacity is influenced by a variety of biological, psychological and sociocultural factors, which influence both the individual and tissue-specific response to biomechanical loading. For example, anatomical factors such as bimalleolar width and calf muscle girth are thought to influence the capacity of the tibia to withstand bending loads (421), while factors such as trochlea dysplasia and patella alta are thought to increase the vulnerability of the patella to lateral displacement (422). Additionally, psychological stressors may impair tissue recovery between loading bouts or influence the fatigue status of an individual prior to subsequent loading bouts (32, 40). Factors such as these may all influence the load capacity of specific tissues, which when combined with sub-optimal kinematic patterns, may increase an individual’s vulnerability to a site-specific injury. Therefore, from both a clinical and a research perspective, it is important to acknowledge the potential interaction between multiple different factors and how this may influence an individual’s vulnerability to running related injuries.

Although the results in this thesis have identified similar kinematic parameters to be associated with different running injuries, there is also likely to be subgroups of runners who demonstrate kinematic patterns beyond those identified in the present study. Kinematic subgroups have previously been reported within specific pathologies such as patellofemoral pain (192 2011) and ITBS (239). For example, amongst runners with patellofemoral pain, Dierks et al (189) reported three distinct kinematic subgroups; one group with increased knee valgus, one with increased hip abduction and one with increased hip internal rotation and knee adduction. Therefore, while the present study identified kinematic characteristics amongst a heterogenous population of injured runners, homogenous subgroups with different kinematic characteristics are also likely to exist. Subsequently for these subgroups, it may be necessary for individually tailored interventions which target the specific kinematic patterns. However, knowledge gained through this thesis, in which kinematic parameters have been associated with multiple injuries irrespective of gender, provides a clear foundation to develop and rigorously test injury prevention interventions which have the potential to offer benefit across multiple subgroups of runners.

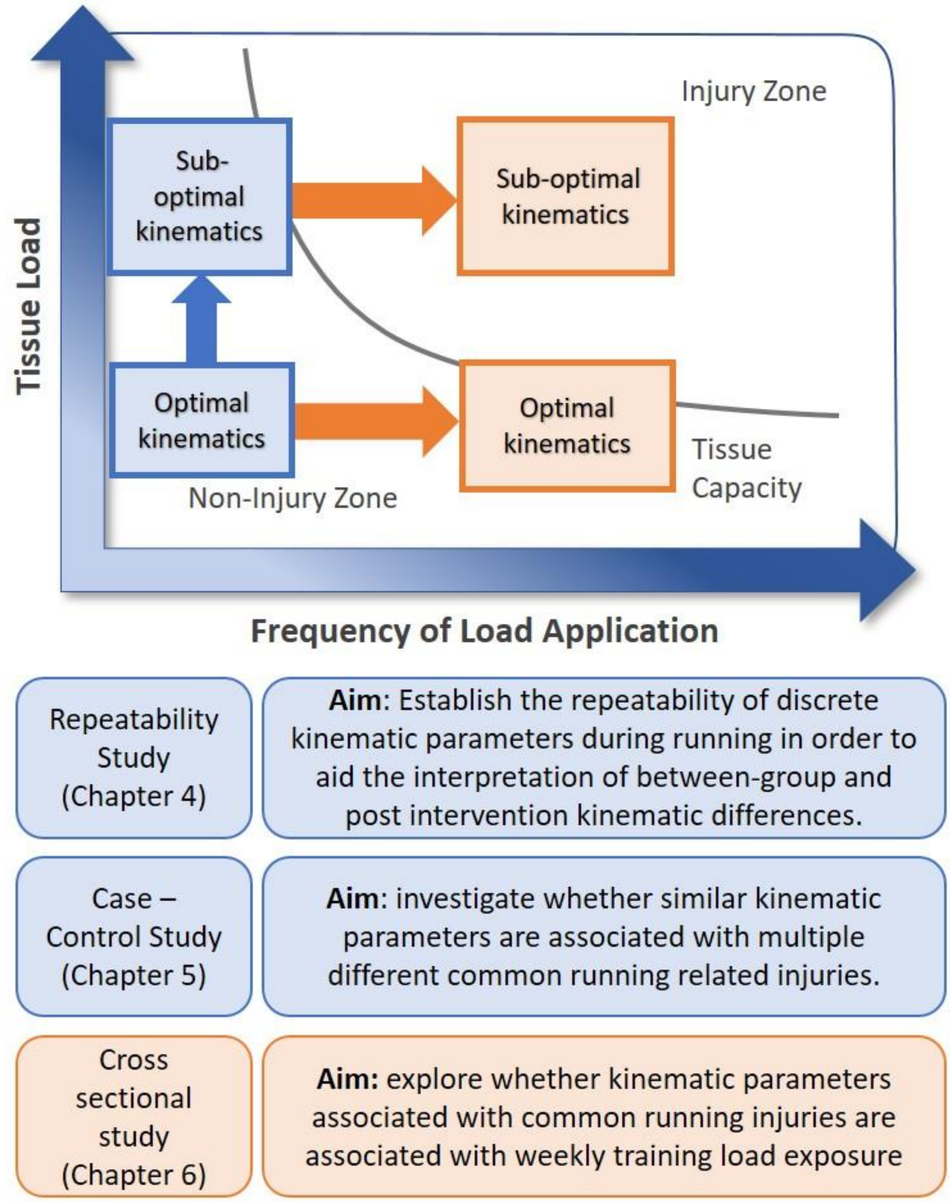
A limitation of the present findings is the retrospective study design, which means causal relationships between the observed kinematics and injury cannot be established. Currently only a limited number of prospective studies have been conducted investigating kinematic contributors to future injury development (54, 57, 59, 149, 151). Without further prospective studies it is possible that the observed kinematic patterns could represent biomechanical adaptions to injury, rather than the cause of injury itself. However, inconsistent evidence currently exists to suggest kinematics adapt in response to injury. Although Fox et al (191) reported differences in kinematic patterns between runners with acute and chronic PFP, Noehren et al (193) found that runners with PFP did not alter their hip kinematics in response to pain. Additionally, findings from retrospective and prospective studies of runners with ITBS have reported similar kinematic patterns. In a retrospective study by Ferber et al, (26) female runners with ITBS were found to have increased hip adduction and knee internal rotation when compared to controls, which is a similar finding to Noehren's (59) prospective study of female runners who developed ITBS. Considering the mechanistic link between kinematics and tissue stress (Chapter 2, Section 2.1; Chapter 5, Section 5.4), it is possible that kinematic patterns could have been present prior to injury development. However further prospective studies which include kinematic data collection following injury development are now required to investigate whether kinematic patterns, such as CPD, are indeed the cause of injury, or adaptations to injury.

Although the kinematic parameter of peak CPD was identified to be associated with multiple common running injuries, it is important to note that running injury development is more complex than simply possessing an injury risk factor. Instead, injury development is the result of a complex interaction between multiple risk factors, exposures and the psychological, biological and behavioural response of the individual. Although identifying singular factors associated with injury may facilitate the development of targeted interventions for effective injury prevention strategies, there needs to be a greater understanding of the interactions between risk factors and exposures which ultimately influence injury aetiology (19, 28, 32, 396).

One such interaction may exist between running kinematics and external training load exposure. As proposed within the introduction, running kinematics are thought to increase tissue load per foot contact of a run. However, without an exposure to external training load, the cumulative tissue load encountered may be insufficient to influence injury development (28, 396). Despite the theoretical interaction between kinematics and training load exposure, Chapter 2 identified limited evidence exploring whether an interaction between training load exposure and kinematics exists. Subsequently, Chapter 6 aimed to explore this gap in the evidence, by investigating whether kinematic parameters associated with common running injuries are associated with weekly training load exposure.

Figure 38 highlights the potential knowledge gap which Chapter 6 aimed to explore. Through understanding whether kinematic differences exist between injury-free high-mileage and low-mileage runners this may provide insight into whether kinematics adaptations, if any, are required to attain regular high-volume training loads while remaining injury free. This may also provide an initial theoretical understanding as to why some runners become injured as training volume increases, while others do not.

Figure 38: Using the adapted stress frequency curve from Hreljac & Ferber (1), the Figure provides an illustration of the knowledge gaps which relevant chapters of the thesis aimed to address. Blue boxes represent the aims achieved and the contribution to knowledge provided. Orange boxes represent the aims of the specific chapters and the specific knowledge gap targeted through fulfilling these aims.



8.4 The influence of external training volume on running kinematics.

The main finding of Chapter 6 was that the prevalence of "high risk" CPD angles differed amongst populations of injury-free high-mileage and low-mileage runners. Specifically, there was a significantly lower prevalence of "high risk" CPD angles amongst injury-free high-mileage runners. Although the cross-sectional nature of this study means that causal relationships cannot be established, we suggest these findings provide preliminary evidence for the existence of a complex interaction between kinematics and training load exposure. This idea supports recent theoretical running injury frameworks which suggest injury causation extends beyond simply possessing a risk factor for injury (28, 33, 396).

One possibility is that the external training loads attainable by runners, could be influenced by their baseline running kinematics. This may explain the lower prevalence of “high-risk” running kinematics amongst injury-free high-mileage runners. This is because the combined tissue loads imposed through sub-optimal kinematics and frequency of load exposure could result in a cumulative tissue load that exceeds tissue capacity. Subsequently in order to attain regular high mileage running, and remain injury free, it may be necessary for runners to either adapt aspects of their gait, or inherently possess kinematic features that minimise the load placed upon the musculoskeletal system.

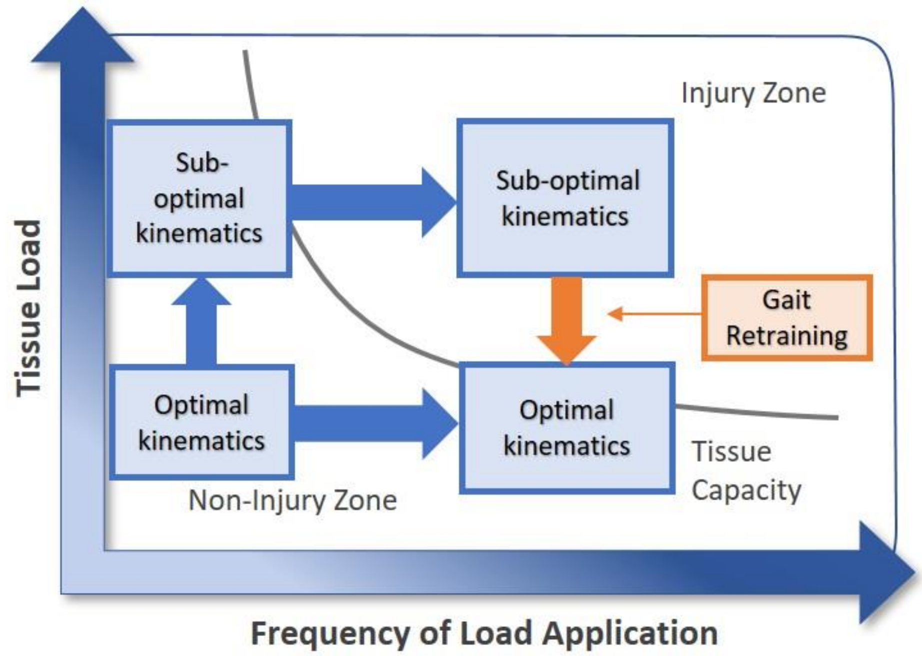
The present findings may have implications for load monitoring amongst runners. According to a recent IOC consensus statement, the accurate monitoring of load exposure is essential for the successful management and prevention of injury in sport (32). However, at present there is conflicting evidence as to whether training load errors contribute to running related injuries (23, 42), which subsequently limits practical recommendations for the safe progression of training volume. One explanation is that current training load measurements may not accurately reflect the cumulative tissue load encountered during running.

Amongst the running literature, current methods of load monitoring predominantly utilise external metrics which quantify the training “dose”, such as volume or duration of running. Although these metrics provide a measure of the external load application, they do not consider factors influencing tissue specific load per stride, such as running kinematic patterns. Considering kinematics influence tissue load per stride and the external training “dose” influences the frequency of load application, the interaction between the two may influence the cumulative tissue load encountered. Consequently, future studies should consider methods of training load monitoring which more accurately reflect the cumulative load encountered during running.

One area for future research would be to consider kinematics as an effect measure-modifier, where the effect of training load upon injury incidence is modified by the kinematic features an individual possesses. Stratifying groups of runners according to their baseline running kinematics would permit the investigation as to whether the maximal training loads attainable by runners, are influenced by their baseline kinematics. Alternatively, with the ongoing development of wearable technology, it may be possible for future studies to monitor in-field training loads which also consider the tissue loads imposed through running kinematics. By utilising such methods, this may help identify specific training-load thresholds in which kinematic patterns will begin to become an important influence on injury development. Through prospective study designs which incorporate graded load exposure, this could also allow for the investigation of whether runners do indeed adapt aspects of their gait. The benefit of such knowledge is that it would assist in the development of practical guidelines for training recommendations or emphasise the need for preventative gait retraining programs to aid the safe progression of running volume.

It is important to note that simply monitoring external training volume and running kinematics does not capture the true training stress encountered by the individual. Kinematics and training volume represent just two singular factors influencing tissue loads, and therefore does not take into account additional biological, psychological and environmental factors which influence load application and the stress response of the individual. Amongst other sports, the use of internal training load metrics has been proposed as a method reflecting the physiological response to load, capturing both biological and psychological contributors to load application. Using session rating of perceived exertion, studies have reported strong associations with future injury development (270). However, at present, this has not been incorporated within running populations. Therefore, future studies should consider methods which attempt to capture the interactions between factors influencing external tissue load application and the internal tissue stress, while still accounting for the effect of individual components. One approach could be to utilise internal training load metrics while stratifying runners according to their baseline kinematics, as this may provide a more accurate reflection of the training stress encountered by runners; representing the physiological stress for a given external load exposure, while acknowledging the biomechanical contributions to load application.

Figure 39: Using the adapted stress frequency curve from Hreljac & Ferber (1), the Figure provides an illustration of the knowledge gaps which relevant chapters of the thesis aimed to address. Blue boxes represent the aims achieved and the contribution to knowledge provided. Orange boxes represent the aims of the specific chapters and the specific knowledge gap targeted through fulfilling these aims



Repeatability Study (Chapter 4)	**Aim**: Establish the repeatability of discrete kinematic parameters during running in order to aid the interpretation of between-group and post intervention kinematic differences.
Case – Control Study (Chapter 5)	**Aim**: investigate whether similar kinematic parameters are associated with multiple different common running related injuries.
Cross sectional study (Chapter 6)	**Aim:** explore whether kinematic parameters associated with common running injuries are associated with weekly training load exposure
Case- Series study (Chapter 7)	**Aim:** investigate whether a simple method of gait retraining can be used to improve biomechanics, clinical and functional outcomes amongst injured runners.

8.5 Gait retraining: targeting running kinematics in injured populations.

Having identified kinematic parameters associated with multiple different running related injuries (Chapter 5), and investigating their association with training load exposure (Chapter 6), Chapter 7 sought to investigate whether gait retraining can be used to improve biomechanics, clinical and functional outcomes amongst injured runners with sub-optimal kinematics at baseline (Figure 39). The impact of achieving this aim, was to provide preliminary evidence for the clinical effectiveness of a simple method of gait retraining amongst runners with PFP which can be easily integrated into clinical practise. Subsequently, amongst runners who possess kinematic patterns which increase tissue loading and have reached a cumulative tissue load which exceeds tissue capacity, gait interventions which target these parameters may serve to reduce tissue load, allowing runners to function within their limits of tissue capacity (Figure 39).

Using a population of runners with PFP, Chapter 7 found a 10% increase in step rate to result in significantly reduced peak contralateral pelvic drop and hip adduction. This coincided with significantly reduced worst pain experienced in the last week as well as improvements in weekly running volume and longest distance run pain free. These findings achieve the final overarching aim of this thesis, which was to investigate whether gait retraining can be used to effectively improve biomechanics, clinical and functional outcomes amongst injured runners. Considering that kinematic patterns may influence tissue load per foot contact of a run, gait retraining interventions which successfully target kinematic patterns associated with injury, could reduce tissue load, offering a potential mechanical explanation for the observed improvements in clinical outcomes.

In contrast to previous step rate interventions, we targeted the gait retraining to injured runners who demonstrate sub-optimal kinematic patterns at baseline. Recent expert review articles have recommended tailoring interventions to the individual patients' deficits in order to optimise clinical outcomes in PFP (423, 424). Despite this recommendation, previous step rate intervention studies have not utilised such an inclusion criterion, instead more broadly including participants with patellofemoral pain (255, 267, 318). Consequently, in many previous studies participants continued to report their worst pain experience in the last week to be 3 out of 10 or greater using the NRS. In contrast, participants within the present study reported an average worst pain of 0.3 out of 10. By recruiting only injured participants who demonstrated kinematics in a region similar to those associated with injury in Chapter 5, we feel this may have reduced the potential for the inclusion of biomechanical non-responders which may have occurred within other studies. This supports the recommendation, proposed by review articles, that interventions should be tailored to patient's deficits (423, 424).

The observed improvements in kinematics and clinical outcomes also coincided with an increase in weekly training exposure, with participants on average returning to their pre-injury weekly training volume (pre-injury: 29.03km, post retraining: 28.33km). Previous studies have reported that targeting running kinematics through gait retraining may serve to reduce tissue load per stride and ultimately the cumulative load across a given run. For example, following a 10% reduction in stride length, Willson et al (425) reported a 17% reduction in peak patellofemoral joint reaction force per stride and a 20% reduction per kilometre. In the context of the present findings, the reductions in peak CPD and hip adduction, may have served to reduce patellofemoral load per stride and ultimately the cumulative patellofemoral load per run. Subsequently by reducing cumulative tissue load, this may allow runners to increase training volume while functioning within the limits of their tissue capacity (Figure 39).

These findings, along with the findings of the thesis, may offer implications for rehabilitation and injury prevention strategies targeted to multiple different running injuries. Within Chapter 5, similar kinematic parameters were observed across multiple different running related injuries. These parameters included peak CPD, increased trunk forward lean and an extended knee and dorsiflexed ankle at initial contact. Apart from forward trunk lean, these parameters have all been reported to be successfully targeted through increasing step rate (Chapter 2, Table 20). The results from the present study highlight that targeting kinematic parameters through gait retraining, may serve to reduce tissue load per foot contact resulting in positive clinical outcomes. Although these results are limited to runners with PFP, the similarities in mechanics associated with different injuries suggests that gait retraining could prove beneficial across multiple different running injuries. Therefore, future studies should consider investigating the clinical effectiveness of gait retraining interventions in additional running related injuries.

The identification of kinematic parameters associated with multiple different running injuries (Chapter 5) and the successful clinical outcomes utilising gait retraining interventions (Chapter 7), could subsequently be used to inform injury prevention interventions targeted towards multiple different running related injuries. Emerging evidence is suggesting that gait retraining interventions, delivered as part of injury prevention programs, may successfully reduce future injury incidence (399). In a randomised control trial with a one year follow up, Chan et al (399) reported a significantly lower injury incidence amongst runners who received a baseline gait retraining intervention targeted at impact loading rates. However, this intervention encouraged runners to land with a forefoot strike pattern which is known to increase lower limb load at the ankle. Consequently, although a lower overall injury rate was observed in the retraining group, there was an increase in calf and Achilles injuries. In contrast, limited adverse effects have been reported when increasing running step rate (399), therefore it is possible that step rate retraining may prove a clinically beneficial intervention in future preventative studies.

A novel finding of Chapter 7 was that runners could self-administer retraining sessions outside of a laboratory and clinical setting. Many current gait retraining interventions are restricted to clinical or laboratory settings, using faded feedback designs or real time feedback procedures (69, 70, 255, 399). Consequently, the time and equipment requirements may not be feasible in many clinical environments which may limit the clinical integration of gait retraining. The present findings provide preliminary evidence for a simple intervention that can be easily integrated into clinical settings and a runner's routine. However, an ongoing limitation remains the ability to identify patients who may benefit from gait retraining interventions and identify biomechanical changes without the use of 3D kinematic measurement systems. Therefore, further work is now required to develop kinematic assessment methods that can be easily integrated into clinical practice, which produce valid and reliable results when compared to 3D methods. In doing so, this would potentially facilitate the more widespread use of gait analysis assessments and gait retraining interventions in clinical environments.

Although the step rate intervention is proposed to address mechanical deficits, without a control group we cannot rule out potential treatment effects extending beyond those that are purely biomechanical. Psychological factors such as increased anxiety, depression and fear of movement or reinjury are known to contribute to the persistence of PFP symptoms and have negative impact upon return to sport outcomes (79, 416, 423, 426). It is possible that the nature of the intervention provided may have indirectly addressed possible psychological barriers to recovery. Through instructing participants to self-administer retraining sessions and self-progress their training volume, this may have facilitated an internal locus of control, promoted active coping strategies and improved self-efficacy. This may have subsequently led to behavioural responses such as the graded progression of training loads and reintegration with social peer groups, leading to positive biological and cognitive processes influencing the clinical and functional outcomes observed. These factors may have ultimately influenced clinical outcomes observed. To provide further insight into the factors, randomised control trials are necessary to establish underlying mechanisms facilitating clinical improvements. We would recommend that future studies also consider investigating the potential psychological impact of gait retraining interventions. Utilising psychological scales at baseline, such as fear avoidance beliefs questionnaire or the Tampa scale for kinesiophobia, it would be possible to quantify the psychological influence of gait retraining interventions upon patient outcomes (423, 427).

8.6 Conclusion

The overarching aim of this thesis was to identify biomechanical characteristics associated with common running injuries, explore whether training load exposure influences running kinematics and finally, investigate whether gait retraining can be used to improve biomechanics, clinical and functional outcomes amongst injured runners. The findings suggest that similar kinematic parameters may underlying multiple different running related injuries, increasing the load placed upon the musculoskeletal system during each stride of a run. Interestingly, these kinematic patterns are less frequently observed amongst injury-free high mileage runners. This suggests an interaction between tissue load imposed through running kinematics and the frequency of load application associated with high mileage training, which may result in a cumulative tissue load that may not be sustainable at during high-volume training. Future prospective studies are now required to investigate whether the kinematics observed are the cause or result of injury and whether runners adapt their kinematics or become injured as training loads increase.

The findings of the thesis also provide preliminary evidence to support the clinical effectiveness of a simple, clinically applicable gait retraining intervention, through a self-administered 10% increase in running step rate. This intervention effectively targeted the kinematic parameters of contralateral pelvic drop and hip adduction which were observed to be associated with common running injuries. Although the present study targeted runners with PFP, the association between contralateral pelvic drop and multiple different running injuries suggests increasing step rate could be beneficial in the rehabilitation process of multiple different running injuries. By reducing tissue load per stride, this approach may facilitate a gradual increase in external training load. Future randomised control trials are now required to investigate the effect of increasing step rate across multiple different running related injuries.

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10 Appendix A: Ethical Approval

10.1 HSCR13/17a

University of Salford MANCHESTER

Research, Innovation and Academic Engagement Ethical Approval Panel

College of Health & Social Care
AD 101 Allerton Building
University of Salford
M6 6PU

T +44(0)161 295 7016
r.shuttleworth@salford.ac.uk

www.salford.ac.uk/

12 June 2013

Dear Chris,

RE: ETHICS APPLICATION HSCR13/17 – 3D kinematics of the trunk pelvis and lower limbs during running

Following your responses to the Panel's queries, based on the information you provided, I am pleased to inform you that application HSCR13/17 has now been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible.

Yours sincerely,

Rachel Shuttleworth

Rachel Shuttleworth
College Support Officer (R&I)

10.2 HSCR13/17b

Research, Innovation and Academic Engagement Ethical Approval Panel

Research Centres Support Team
G0.3 Joule House
University of Salford
M5 4WT

T +44(0)161 295 2280

www.salford.ac.uk/

23 November 2015

Dear Steve,

RE: ETHICS APPLICATION HSCR 13-17 – 3D kinematics of the trunk pelvis and lower limbs during running

Based on the information you provided, I am pleased to inform you that your request to amend application HSCR13-17 has been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible by contacting Health-ResearchEthics@salford.ac.uk

Yours sincerely,

Sue McAndrew
Chair of the Research Ethics Panel

10.3 HSCR16-49

University of Salford MANCHESTER

Research, Innovation and Academic Engagement Ethical Approval Panel

Research Centres Support Team
G0.3 Joule House
University of Salford
M5 4WT

T +44(0)161 295 2280

www.salford.ac.uk/

5 July 2016

Dear Chris,

RE: ETHICS APPLICATION HSCR 16-49 – Do simple running technique changes reduce pain and change injury causing mechanics in runners with common running related injuries?

Based on the information you provided, I am pleased to inform you that application HSCR16-49 has been approved.

If there are any changes to the project and/ or its methodology, please inform the Panel as soon as possible by contacting Health-ResearchEthics@salford.ac.uk

Yours sincerely,

Sue McAndrew
Chair of the Research Ethics Panel

11 Appendix B: Participant Information Sheet

11.1 Information sheet: Injured & control subjects

Participant information sheet (Version 1 02/03/2013)

University of Salford MANCHESTER

Running Biomechanics Assessment

A comparison of biomechanics during running and its relationship to clinical assessment measures.

School of Health, Sports & Rehabilitation Sciences
Researcher: Christopher Bramah & Dr Steve Preece

Thank you for volunteering to take part in our research project analysing the biomechanics of running. Before you decide to take part in the study it is important for you to understand why the research is being done and what will e involved in the process. Below are details of the biomechanics analysis procedure that will take place when you arrive including the purpose of the project, the risks and the benefits of participating. Should you have any questions or require any further information please do not hesitate to contact myself via the provided email address.

Background to the Study

Many running related injuries have been reported to be caused by multiple factors including altered biomechanics. If altered biomechanics are sustained over a long period of time we think this may result in running injuries. When people sustain a running injury they often see a physiotherapist who conducts a physical assessment in order to determine the cause of the injury. We want to investigate the running biomechanics of the spine, pelvis and lower limbs in people of different running abilities in order to identify if there are any differences between runners of different abilities and to see if physical assessment measures used in clinical practise relate to normal running biomechanics.

Participating in this study is completely voluntary and you may withdraw at any time. You will not be penalised if you decline to participate or if you choose to withdraw at any time.

Procedure

On arrival to the gait analysis clinic you will be met by one of our research team. We will then complete consent forms, a running history form detailing your times achieved for 5km, 10km, injury history and current levels of running, as well as a medical questionnaire to ensure there is no risk to your health during participation. We will then attach reflective markers to your chest, lower back, hips, thighs, lower legs and shoes. These markers will allow our system to record how different joints of the body move when you run.

You will then be given 4 to 5 minutes running to warm up using the treadmill which will help to make sure you are comfortable and running naturally. You will then run on the treadmill at 4 different running speeds for 2 minutes starting at a slow jogging pace and gradually increasing the speed to a running pace similar to your 5km running speed. Between running trials you will be permitted a rest period of up to 2 minutes between each trial. While you run we will record your biomechanics using high speed cameras.

Once your biomechanics have been recorded a qualified physiotherapist will take measurements of your joint movements (like your ankle movement) and strength of different joints. To conduct the tests the physiotherapist will move your joints including your ankle, knee and hips in order to measure how much they move. The physiotherapist will then take some strength tests by asking you to use muscles to produce a force against a resistance. This will help to inform our assessment of your biomechanical patterns. The total testing procedure will take approximately 2 hours with running time a maximum of 13 minutes with rest periods in between.

Risks & Benefits

As you are required to run on a treadmill there is a risk of developing an injury during the procedure. In order to reduce this risk we will give you time to warm up and adapt to running on the treadmill. You will have as long as you require to warm up and we will only begin the testing once you say you are adequately warm and ready to run.

Following the test you will be provided with a report comparing your biomechanics to a database of elite athletes and comparing the symmetry of your movement between right and left. This report will detail any differences identified in the assessment procedure.

Confidentiality of Subject Details

All details and data collected will be coded using a subject number and stored on a password protected hard drive to ensure that your personal information is kept anonymous. At any point should you decide you no longer wish to take part in our study then you can withdraw and your data will be destroyed. Should you have any questions following completion of the study you can again contact the lead researcher via the email address provided on this letter.

Important information

If you have a history of, and or currently have any injuries, illnesses or medical conditions, take any medication or have any prescribed medications then you must inform ourselves prior to taking part in our research.

Participant information sheet (Version 1 02/03/2013)

Things to bring

Comfortable running shorts, your running trainers and a running crop top if you are female.

Contact details

If you have any questions about your participation, require any further information or have decided you no longer wish to take part in this study, please contact myself via email at c.bramah1@edu.salford.ac.uk

Thank you for reading this document

Christopher Bramah

Post graduate researcher
University of Salford

11.2 Information sheet: Gait retraining

Participant information sheet (Version 2 24/06/2016)

University of Salford MANCHESTER

Running Biomechanics Assessment

The effect of running technique changes on pain and running biomechanics

School of Health, Sports & Rehabilitation Sciences
Researcher:

Thank you for volunteering to take part in our research project analysing the biomechanics of running. Before you decide to take part in the study it is important for you to understand why the research is being done and what will be involved in the process. Below are details of the biomechanics analysis procedure that will take place when you arrive including the purpose of the project, the risks and the benefits of participating. Should you have any questions or require any further information please do not hesitate to contact myself via the provided email address.

Background to the Study

Many running related injuries have been reported to be caused by multiple factors including altered biomechanics. If altered biomechanics are sustained over a long period of time we think this may result in running injuries.

Simple technique changes during running may change the biomechanics which are proposed to cause common running injuries. The injuries which we are looking at include patellofemoral pain syndrome, iliotibial band syndrome, medial tibial stress syndrome and achilles tendonpathy. We want to investigate whether simple running technique changes can reduce pain and change biomechanics in runners with a current running injury and whether these changes can be maintained.

Participating in this study is completely voluntary and you may withdraw at any time. You will not be penalised if you decline to participate or if you choose to withdraw at any time. If you agree to participate you will be asked to attend the running clinic at two separate occasions approximately 4 weeks apart lasting a maximum of 2 hours.

Procedure

On arrival to the gait analysis clinic you will be met by one of our research team. We will then complete consent forms, questionnaires used to monitor your injury progression, a running history form detailing your times achieved for 5km, 10km, injury history and current levels of running, as well as a medical questionnaire to ensure there is no risk to your health during participation. You will then be assessed by a qualified physiotherapist to confirm the diagnosis of your current injury and ensure you are not at risk of further harm. We will then attach reflective markers to your chest, lower back, hips, thighs, lower legs and shoes. These markers will allow our system to record how different joints of the body move when you run.

You will then be given 4 to 5 minutes running to warm up using the treadmill which will help to make sure you are comfortable and running naturally. You will then run on the treadmill at 2 different running speeds for 2 minutes starting at a slow jogging pace and gradually increasing the speed to a running pace similar to your 5km running speed. Between running trials you will be permitted a rest period of up to 2 minutes between each trial. While you run we will record your biomechanics using high speed cameras and ask you to rate your pain on a scale of 0 being no pain and 10 being sever pain.

Once you have completed this initial testing period you will be asked to run with your footsteps matching an audible metronome in order to increase your stride rate. You will run for 5 minutes starting at a jogging pace and 5 minutes at a pace similar to 10km running pace at the new step rate. At the end of each 5 minute time period you will be asked to again rate your pain on a scale of 0 to 10. As you run we will record your biomechanics using our high speed cameras. The total testing procedure will take approximately 2 hours with running time a maximum of 20 minutes with rest periods in between.

Once you have completed the testing procedure you will be permitted up to 15 minutes of running on the treadmill to allow you to adapt to the new technique. The 15 minute period is entirely up to you and if you do not wish to practise you do not have to. You will then be asked to continue your current running levels with the new technique. You will be asked to contact the researcher on a weekly basis to ask any questions you may have and to make sure you are not running in any worsening pain.

After 4 weeks you will be asked to return to the clinic for a follow up test where you will be asked to complete the same testing procedures motioned above, including completion of questionnaires and running for a maximum of 9 minutes; a warm up period of 4 to 5 minutes and then a further 2 minutes running at two different speeds (a jogging speed and a 10km pace speed). Following this test you will be asked to continue your normal running and a follow up email and questionnaire will be sent to you at a 3 month period to monitor your progress.

Risks & Benefits

As you are required to run on a treadmill there is a risk of developing an injury during the procedure. In order to reduce this risk we will give you time to warm up and adapt to

Participant information sheet (Version 2 24/06/2016)

running on the treadmill. You will have as long as you require to warm up and we will only begin the testing once you say you are adequately warm and ready to run.

There is also a risk that you may experience some pain when you run due to your current injury. A qualified physiotherapist will be there to monitor your pain and you can stop the assessment at any time you wish to do so. You will be also provided the contact details for the researcher and can contact them weekly and at any point should you have any questions.

Confidentiality of Subject Details
All details and data collected will be coded using a subject number and stored on a password protected hard drive to ensure that your personal information is kept anonymous. At any point should you decide you no longer wish to take part in our study then you can withdraw and your data will be destroyed. Should you have any questions following completion of the study you can again contact the lead researcher via the email address provided on this letter.

Important information
If you have a history of, and or currently have any injuries, illnesses or medical conditions, take any medication or have any prescribed medications then you must inform ourselves prior to taking part in our research.

Complaints/ Problems
If you have any complaints about the research then you may discuss these with our lead researcher [contact details for lead researcher here]. However if you do not wish to discuss these issues with the researcher and would like to submit a formal complaint then please contact Anish Kurien using the below contact details:

Anish Kurien Research Centre's Manager

University of Salford

G.08, Joule House, Acton Square, Salford, M5 4WT

t: +44 (0) 161 295 5276 | **e:** a.kurien@salford.ac.uk

Things to bring
Comfortable running shorts, your running trainers and a running crop top if you are female.

Contact details & How to find us
If you have any questions about your participation, require any further information or have decided you no longer wish to take part in this study, please contact myself via email at c.a.bramah1@edu.salford.ac.uk.

Thank you for reading this document

Chris Bramah

Post graduate researcher
University of Salford

12 Appendix C: Consent forms

12.1 Consent form: Injured and Control Participants

Running biomechanics

CONSENT FORM

Version 2 07/02/2014

Title of Project:

A comparison of biomechanics during running and its relationship to clinical assessment measures.

Name of Researcher: Christopher Bramah

Please initial each point

1. I confirm that I have read and understand the information sheet dated......................for the above study.
2. I understand that my participation is voluntary and that I am free to withdraw at any time without my medical care or legal rights being affected
3. I understand that all data collected with be stored securely and anonymously so no personal details can be identified.
4. I understand the (specified) physical requirements of the study and know of no medical reason why I should not participate
5. I understand the risks of using a treadmill including falling, tripping, injury or malfunction of electrical equipment and agree to participate
6. I agree to be videoed while running, for this video to exclude facial features to ensure video footage is anonymous.
7. I agree for all anonymous data collected to be used for research/presentation/ dissemination work.

Name of subject	Subject No.	Date	Signature

(Supervising adult to complete this section if client s under the age of 18)

Researcher	Date	Signature

12.2 Consent form: Gait Retraining

Running biomechanics

CONSENT FORM

Version 2 24/06/2016

Title of Project:

Do simple running technique changes reduce pain and change injury causing mechanics in runners with common running related injuries?

Name of Researcher:

Please initial each point

1. I confirm that I have read and understand the information sheet titled **Participant Information Sheet v2 24/06/2016** for the above study.
2. I understand that my participation is voluntary and that I am free to withdraw at any time without my medical care or legal rights being affected
3. I understand that all data collected with be stored securely and anonymously so no personal details can be identified.
4. I understand the (specified) physical requirements of the study and know of no medical reason why I should not participate
5. I understand the risks of using a treadmill including falling, tripping, injury or malfunction of electrical equipment and agree to participate
6. I agree to be videoed while running, for this video to exclude facial features to ensure video footage is anonymous.
7. I agree for all anonymous data collected to be used for research/presentation/ dissemination work.

Name of subject	Subject No.	Date	Signature
Researcher		Date	Signature

13 Appendix D: Inclusion/ Exclusion Criteria

All Subjects: Exclusion Criteria

Subjective **Exclude if yes to any one of the following:**	
History of musculoskeletal surgery History of traumatic knee dislocation Neurological symptoms affecting gait	Yes No
Objective **Exclude if positive tests for any one of the following:**	
Leg length discrepancy: ASIS to Medial Maleoli >0.5cm **Hip:** Impingement Signs: FABERS FAIR **Knee Meniscus:** McMurrys Grind test Apleys Grind Test **Knee Ligaments:** Varus stress test/ Valgus stress test Lachmans Anterior draw test Posterior Draw test **Ankle:** Posterior Impingement Signs **Tibia:** Shin Oedema Compression of Tibial Body	Positive Negative

Patellofemoral Pain: Inclusion/ Exclusion

Inclusion **(must report yes to one or more item within each of the following sections)**	
History Insidious or gradual onset during running	**Yes** **No**
Symptoms Peripatella or retropatella pain Pain on squatting Pain with any one of the following activities: Stairs/ Kneeling/ Prolonged sitting/ Jumping	**Yes** **No**
Symptom Severity Minimum Pain Running 3/10 NRS	**Yes** **No**
Symptom Duration Minimum 3 month history	**Yes** **No**
Objective Pain with any one of the following: Patella compression Patella apprehension (Clarkes test) Palpation lateral patella facet Isometric quadricep contraction (30 knee flexion)	**Yes** **No**
Exclusion **Exclude if yes to any one of the following:**	
Symptoms Onset following trauma Constant unremitting pain Onset due to participation in any other sporting activity (Also see all exclusion criteria)	**Yes** **No**
Objective (See All Exclusion Criteria)	

Iliotibial Band Syndrome: Inclusion/Exclusion

Inclusion **(must report yes to one or more item within each of the following sections)**	
History Insideous onset during running	**Yes** **No**
Symptoms Lateral Knee Pain Pain eases after cessation of running	**Yes** **No**
Symptom Severity Minimum Pain Running 3/10 NRS	**Yes** **No**
Symptom Duration Minimum 3 month history	**Yes** **No**
Objective Nobles Compression Test Pain on palpation of lateral femoral condyle	**Yes** **No**
Exclusion **Exclude if yes to any one of the following:**	
Symptoms Onset following trauma Constant unremitting pain Onset due to participation in any other sporting activity (Also see all exclusion criteria)	**Yes** **No**
Objective (See All Exclusion Criteria)	

Medial Tibial Stress Syndrome: Inclusion/Exclusion

Inclusion **(must report yes to one or more item within each of the following sections)**	
History Insideous onset during running	**Yes** **No**
Symptoms Distal medial shin pain Pain eases after cessation of running	**Yes** **No**
Symptom Severity Minimum Pain Running 3/10 NRS	**Yes** **No**
Symptom Duration Minimum 3 month history	**Yes** **No**
Objective Shin palpation test: Pain on palpation of the medial ridge of the tibia at the insertion of the tibialis posterior and medial fibres of the soleus	**Yes** **No**
Exclusion **Exclude if yes to any one of the following:**	
Symptoms Onset following trauma Constant unremitting pain Onset due to participation in any other sporting activity (Also see all exclusion criteria)	**Yes** **No**
Objective (See All Exclusion Criteria)	

Achilles Tendinopathy: Inclusion/ Exclusion

Inclusion **(must report yes to one or more item within each of the following sections)**	
History Insideous onset during running	**Yes** **No**
Symptoms Mid portion achilles pain Pain eases into running Morning stiffness easing with movement	**Yes** **No**
Symptom Severity Minimum Pain Running 3/10 NRS	**Yes** **No**
Symptom Duration Minimum 3 month history	**Yes** **No**
Objective Pain on palpation of mid portion of achilles (2cm to 6cm above calcaneus)	**Yes** **No**
Exclusion **Exclude if yes to any one of the following:**	
Symptoms Onset following trauma Constant unremitting pain Onset due to participation in any other sporting activity (Also see all exclusion criteria)	**Yes** **No**
Objective (See All Exclusion Criteria)	

14 Appendix E: Treadmill Accommodation

Methods

A total of 13 injury free participants completed continuous treadmill running for a total of 10 minutes. Thirty seconds of kinematic data were collected at 3 minutes, 5 minutes and 8 minutes during continuous running. All kinematic data was collected and processed in accordance with procedures outlined in section 3.3.

One-way repeated measures ANOVA was used to investigate whether there are significant differences between time points for discrete kinematic parameters of interest with a critical alpha of .05. When significant differences were observed, post hoc Bonferroni testing was used to identify differences between time-points.

Results

Results are presented in the following tables:

	3 minutes	5 minutes	8 minutes	ANOVA p value	Bonferroni Pairwise comparison (P value)	
Stride Rate (Hz)*	1.39 (0.04)	1.41 (0.05)	1.42 (0.05)	<.01*	3min – 5min	.18
					3min – 8min*	<.01*
					5min – 8min	.47
Stride Length (m)*	2.19 (0.06)	2.17 (0.08)	2.16 (0.08)	<.01*	3min – 5min	.23
					3min – 8min*	<.01*
					5min – 8min	.45
Stance Time (sec)	0.49 (0.05)	0.49 (0.04)	0.49 (0.04)	0.24	3min – 5min	.30
					3min – 8min	1.0
					5min – 8min	.39

36 *Spatiotemporal parameters across separate timepoints. Values represent mean (standard deviation). *indicates statistically significant difference at p <.05*

	3 minutes	5 minutes	8 minutes	ANOVA P value
Trunk Forward Lean	3.0 (2.2)	3.6 (2.3)	3.04 (2.19)	.24
Anterior Pelvic Tilt	6.8 (1.9)	7.0 (2.1)	7.0 (2.2)	.57
Hip Flexion	23.3 (4.0)	24.1 (3.8)	23.4 (4.3)	.29
Knee Flexion	10.2 (4.5)	10.8 (5.5)	9.7 (5.0)	.46
Ankle Dorsiflexion	2.7 (5.8)	3.1 (5.4)	2.8 (5.2)	.68

***37**: Kinematic parameters at initial contact. Values represent mean (standard deviation). *indicates statistically significant difference at p <.05*

	3 minutes	5 minutes	8 minutes	ANOVA (P value)	Bonferroni Pairwise comparison (P value)	
Trunk Forward Lean	8.1 (3.5)	8.4 (3.8)	8.5 (3.7)	0.42	3min – 5min	.65
					3min – 8min	.98
					5min – 8min	1.0
Contralateral Pelvic Drop	5.3 (2.6)	4.9 (2.9)	4.9 (2.8)	0.28	3min – 5min	1.0
					3min – 8min	.57
					5min – 8min	1.0
Hip Adduction	12.2 (3.4)	11.9 (3.4)	12.0 (3.5)	0.59	3min – 5min	1.0
					3min – 8min	1.0
					5min – 8min	1.0
Knee Abduction	2.1 (4.2)	2.1 (4.4)	1.8 (4.3)	0.39	3min – 5min	1.0
					3min – 8min	1.0
					5min – 8min	.19
Knee Flexion	33.5 (3.9)	32.8 (4.2)	32.1 (3.8)	<.01*	3min – 5min*	.04*
					3min – 8min*	<.01*
					5min – 8min	.14
Ankle Dorsiflexion	21.2 (2.8)	20.8 (2.9)	20.2 (2.4)	<.01*	3min – 5min*	<.01*
					3min – 8min*	<.01*
					5min – 8min	.26
Rearfoot Eversion	2.9 (5.7)	2.9 (5.7)	5.0 (6.3)	0.32	3min – 5min	1.0
					3min – 8min	.99
					5min – 8min	.92

*38: Peak joint angles during stance phase. Values represent mean (standard deviation). *indicates statistically significant difference at p <.05*

15 Appendix F: Peer reviewed publication – Is there a pathological running gait associated with common soft tissue running injuries? *American Journal of Sports Medicine.*

Check for updates

Is There a Pathological Gait Associated With Common Soft Tissue Running Injuries?

Christopher Bramah,*† MSc, MCSP, Stephen J. Preece,† PhD,
Niamh Gill,† PhD, and Lee Herrington,† PhD, MCSP
Investigation performed at the School of Health Sciences, University of Salford, Salford, UK

Background: Previous research has demonstrated clear associations between specific running injuries and patterns of lower limb kinematics. However, there has been minimal research investigating whether the same kinematic patterns could underlie multiple different soft tissue running injuries. If they do, such kinematic patterns could be considered global contributors to running injuries.

Hypothesis: Injured runners will demonstrate differences in running kinematics when compared with injury-free controls. These kinematic patterns will be consistent among injured subgroups.

Study Design: Controlled laboratory study.

Methods: We studied 72 injured runners and 36 healthy controls. The injured group contained 4 subgroups of runners with either patellofemoral pain, iliotibial band syndrome, medial tibial stress syndrome, or Achilles tendinopathy (n = 18 each). Three-dimensional running kinematics were compared between injured and healthy runners and then between the 4 injured subgroups. A logistic regression model was used to determine which parameters could be used to identify injured runners.

Results: The injured runners demonstrated greater contralateral pelvic drop (CPD) and forward trunk lean at midstance and a more extended knee and dorsiflexed ankle at initial contact. The subgroup analysis of variance found that these kinematic patterns were consistent across each of the 4 injured subgroups. CPD was found to be the most important variable predicting the classification of participants as healthy or injured. Importantly, for every 1° increase in pelvic drop, there was an 80% increase in the odds of being classified as injured.

Conclusion: This study identified a number of global kinematic contributors to common running injuries. In particular, we found injured runners to run with greater peak CPD and trunk forward lean as well as an extended knee and dorsiflexed ankle at initial contact. CPD appears to be the variable most strongly associated with common running-related injuries.

Clinical Relevance: The identified kinematic patterns may prove beneficial for clinicians when assessing for biomechanical contributors to running injuries.

Keywords: running; kinematics; injury; gait

Running is an increasingly popular method of physical activity; however, it also poses a risk of injuries to the musculoskeletal system. It has been reported that approximately 50% of runners become injured annually, with 25% injured at any one time.[13] The majority of running-related injuries are considered to be overuse injuries, with the most frequently injured sites including the knee, foot, and lower leg, with incidence rates reported of around 50%, 39%, and 32%, respectively.[46] Less common injury sites include the ankle and lower back, as well as the hip and pelvis, with incidence rates ranging from 4% to 16%, 5% to 19%, and 3% to 11%, respectively.[46] Of all running-related injuries, the most frequently cited injuries include patellofemoral pain (PFP), iliotibial band syndrome (ITBS), medial tibial stress syndrome (MTSS), Achilles tendinopathy (AT), plantar fasciitis, stress fractures, and muscle strains.[24,44] Many of these injuries are known to have high recurrence rates, leading to a reduction or cessation of training in approximately 30% to 90% of cases.[47] The factors related to the development of running-related injuries are multifactorial and diverse; however, it is widely accepted that abnormal running kinematics play a role.[1,7,31]

There has been a large amount of research that has sought to identify the kinematic patterns associated with

*Address correspondence to Christopher Bramah, MSc, MCSP, School of Health Sciences, University of Salford, Blatchford Building, Salford, M6 6PU, UK (email: c.a.bramah@salford.ac.uk).

†School of Health Sciences, University of Salford, Salford, UK.

The authors declared that they have no conflicts of interest in the authorship and publication of this contribution. AOSSM checks author disclosures against the Open Payments Database (OPD). AOSSM has not conducted an independent investigation on the OPD and disclaims any liability or responsibility relating thereto.

The American Journal of Sports Medicine
1–9
DOI: 10.1177/0363546518793657

many common soft tissue running injuries, including MTSS,[26] PFP,[32,52] ITBS,[12,31] and AT.[39] Interestingly, many of these studies have reported similar kinematic patterns to be associated with different running injuries. For example, increased hip adduction has been associated with PFP[32,52] and ITBS,[12,31] and increased hip internal rotation has been associated with PFP[41] and MTSS.[26] Research has also suggested that because of kinematic coupling between the femur, knee, and foot, increased hip adduction or hip internal rotation may contribute to greater rearfoot eversion.[2,27,38] Interestingly, increased rearfoot eversion has been associated with injuries such as MTSS[3,50] and AT.[39] This research suggests that there may be a number of similar kinematic patterns that could underlie multiple different soft tissue running injuries. It is possible that these patterns could lead to elevated stress on multiple anatomic structures, leading to the development of injuries at different areas. These kinematic patterns may represent global contributors to injuries.

Recent research supports the idea of biomechanical parameters that could be considered global contributors to running injuries. In a prospective study of 249 runners, Davis et al[7] reported that runners who went on to develop a range of different injuries demonstrated significantly elevated vertical loading rates. However, in a retrospective study that investigated runners with AT and MTSS, Becker et al[3] reported greater rearfoot eversion in the late stance phase to be a characteristic consistently associated with an injury. Although these 2 studies provide preliminary evidence for the existence of global contributors to running injuries, Davis et al[7] did not include kinematic data, while Becker et al[3] investigated only MTSS and AT. Therefore, further research is required to understand whether there are similar kinematic patterns that may underlie multiple different running injuries. This understanding would be invaluable to clinicians as it could be used as a basis for both screening techniques and preventative and rehabilitative programs.

The aim of this current study was to identify whether there are kinematic parameters that may represent global kinematic contributors to running injuries. To achieve this objective, we sought to identify whether there are differences in running kinematics between a large group of runners with common running injuries (ITBS, PFP, MTSS, and AT) compared with a healthy control group. We hypothesized that the pooled group of injured runners would demonstrate greater contralateral pelvic drop (CPD), hip adduction, and rearfoot eversion when compared with injury-free controls. We also hypothesized that these kinematic patterns would be consistent among injured subgroups.

METHODS

Participants

A total of 108 runners were enrolled in this current study, including 72 injured runners (28 male, 44 female) and 36 healthy controls (15 male, 21 female) matched for age, height, and weight (Table 1). The injured group contained subgroups of 18 runners each with PFP, ITBS, MTSS, and AT (Table 2). These injuries were selected as they are cited as the most prevalent soft tissue overuse running injuries.[24] An a priori sample size calculation was conducted using data from a previous study reporting kinematic differences between healthy and injured runners.[32] Using G*Power software, we calculated that we would need at least 98 people (65 injured) to detect an effect size of 0.75 with a power of 0.85 and a critical alpha of .01. Participants were recruited via poster advertisements at local running clubs and sports injury clinics. All participants provided written informed consent before participation, and ethical approval was obtained via the local ethics committee.

TABLE 1
Participant Characteristics[a]

	Healthy (n = 36)	Injured (n = 72)
Age, y	33.2 ± 8.4	34.8 ± 9.9
Weight, kg	60.8 ± 8.4	63.4 ± 10.5
Height, cm	171.6 ± 7.3	170.7 ± 8.6
Body mass index, kg/m^2	20.6 ± 1.8	21.7 ± 2.7
Miles run per week	60.5 ± 23.2[b]	21.2 ± 13.1[b]

[a]Values are presented as mean ± SD.
[b]Statistical significance at $P \le .01$.

Inclusion/Exclusion Criteria

Injured Group. The injured group included participants with a current diagnosis of either PFP, ITBS, MTSS, or AT. Injury diagnosis was confirmed after a physical examination by a qualified physical therapist in accordance with previously published diagnostic criteria for PFP,[6] ITBS,[17] MTSS,[54] and AT[22] (see Appendix 1, available in the online version of this article). All participants reported being able to run up to 10 minutes before the onset of pain and maximal pain during running of greater than 3 of 10 on a numerical rating scale (0 = no pain, 10 = worst possible pain). Additionally, all participants reported that they were not currently receiving medical treatment for their injury and that their pain had caused a restriction to their running volume and/or frequency for a minimum of 3 months. Previous research has reported training factors such as a rise in weekly training volume to increase the risk of injuries. This is likely because of a sudden excessive rise in acute tissue stress on the musculoskeletal system, resulting in insufficient time for adaptive changes.[33] Therefore, to control for training errors as a cause of injury, participants were excluded if they reported an increase in weekly training volume of greater than 30% preceding the onset of injury.

Healthy Group. Control participants were included if they reported running a minimum of 30 miles per week for the past 18 months with no reported injury. Participants were excluded if they reported any musculoskeletal ailment within the past 18 months that caused a restriction

TABLE 2
Characteristics of Injured Subgroups[a]

	PFP (n = 18)	ITBS (n = 18)	MTSS (n = 18)	AT (n = 18)
Age, y	34.5 ± 9.4	34.3 ± 7.9	31.9 ± 9.7	38.5 ± 11.7
Weight, kg	64.4 ± 9.6	63.6 ± 11.2	62.5 ± 10.1	63.1 ± 11.8
Height, cm	173.5 ± 8.5	170.6 ± 8.5	167.3 ± 8.1	171.6 ± 8.7
Body mass index, kg/m^2	21.3 ± 1.9	21.8 ± 3.3	22.2 ± 2.3	21.3 ± 2.0
Miles run per week	18.6 ± 6.9	14.8 ± 5.8	19.5 ± 12.2	31.9 ± 17.6[b]

[a]Values are presented as mean ± SD. AT, Achilles tendinopathy; ITBS, iliotibial band syndrome; MTSS, medial tibial stress syndrome; PFP, patellofemoral pain.
[b]Statistical significance at $P \leq .01$.

or cessation of running or any need to seek advice from a health care professional. Additional exclusion criteria included a history of overuse running injuries, injuries caused by another sport, previous spinal injuries, or lower limb surgery.

Procedures

Kinematic data were collected from all participants while running on a treadmill at 3.2 m/s wearing their own running shoes. After a 5-minute warm-up period, 30 seconds of kinematic data were collected using a 12-camera Oqus system (240 Hz; Qualisys). A total of 9 anatomic segments were tracked following a previously published protocol by the same authors shown to have good to excellent repeatability.[28,37] Segments included the thorax, pelvis and bilateral thigh, shank, and foot segments. In addition, a further rearfoot segment was included using 3 noncollinear markers attached to the heel of the participant's shoes. The foot segment was used to calculate sagittal plane ankle kinematics, while the rearfoot segment was used to calculate frontal plane foot kinematics. Further details of the markers used to track each segment and the precise definition of the anatomic coordinate systems are provided in Appendix 2 (available online) and described in previous publications.[14,28,37]

Raw kinematic data were low pass filtered at 10 Hz. Intersegmental kinematics, along with the motions of the pelvis and thorax with respect to the laboratory system, were calculated using a 6 degrees of freedom model with the commercial software Visual3D (C-Motion). Gait events were defined using a kinematic approach[20] and subsequently used to segment each kinematic signal into a minimum of 10 consecutive gait cycles. An ensemble average for each signal was determined and selected kinematic parameters derived from the ensemble average curves. This latter processing was carried out using a custom Matlab script (MathWorks).

Data Analysis

A range of kinematic parameters at both initial contact and midstance was selected for analysis. Parameters at initial contact included sagittal plane angles of the trunk, pelvis, hip, knee, and ankle as well as frontal plane angles of the trunk and rearfoot. Peak angles at midstance included sagittal and frontal plane angles of the trunk, pelvis, knee, ankle, and rearfoot as well as transverse plane angles of the hip and knee. Parameters were selected based on previous research reporting differences between injured and noninjured runners.[39,41,52] Peak angles at midstance were defined as the maximum joint angle between initial contact and toe-off. Foot strike patterns of each group were determined based on the kinematic waveforms of the ankle joint. If the ankle demonstrated an immediate movement into plantarflexion, participants were classified as having a rearfoot strike, while participants demonstrating immediate ankle dorsiflexion were classified as having a forefoot strike. The injured leg was analyzed from the injured runners, and the right or left leg was analyzed at random from the healthy runners to match the total distribution of right and left legs in the injured group.

Statistical Analysis

Participant characteristics were analyzed using independent t tests for the healthy versus injured group comparison and 1-way analysis of variance (ANOVA) for the subgroup comparison (Tables 1 and 2). Chi-square tests were used to assess for differences in the distribution of foot strike patterns between the groups. To identify possible global contributors to running injuries, we used a 2-phased approach. First, data from the injured group were pooled and kinematic parameters compared with those of the healthy group using an independent t test. Second, for any variables found to be significantly different after the injured versus healthy comparison, we assessed for subgroup differences between the 4 injured subgroups using 1-way ANOVA and a post hoc least significant difference test. To be considered a global contributor to running injuries, a kinematic parameter was required to be consistent across the different injured subgroups. This ensured that differences observed in the pooled injured data were not the result of large effects in one of the injured subgroups. Before analysis, all kinematic parameters were assessed for homogeneity of variance and normal distribution using the Levene test ($P \geq .05$) and Shapiro-Wilk test ($P \geq .05$). When assumptions were not met, an equivalent nonparametric test was used. To reduce the possibility of

TABLE 3
Kinematic Angles at Initial Contact[a]

	Healthy	Injured	*P* Value	Effect Size
Trunk forward lean, deg	3.9 ± 2.9	5.7 ± 3.9	.033	0.52
Trunk ipsilateral lean, deg	2.5 ± 1.8	3.1 ± 2.2	.257	0.28
Pelvis anterior tilt, deg	5.9 ± 3.3	7.0 ± 3.8	.132	0.32
Knee flexion, deg	10.2 ± 4.8	6.0 ± 4.9	<.01[b]	0.87
Ankle dorsiflexion, deg	2.4 ± 6.5	7.2 ± 6.9	<.01[b]	0.71
Rearfoot inversion, deg	8.7 ± 6.1	6.2 ± 4.5	.018	0.47

[a]Values are presented as mean ± SD unless otherwise specified.
[b]Statistical significance at $P < .01$.

type I errors, a critical alpha of .01 was used for the injured versus healthy comparison. However, we used a critical alpha of .05 for the subgroup ANOVA because of the smaller subgroup sample sizes. This was deemed appropriate given the smaller number of group comparisons and therefore a lower likelihood of type I errors.

In addition to calculating statistical significance for group comparisons, we also calculated effect sizes. For *t* test comparisons, we used Cohen *d* and interpreted an effect size of 0.2, 0.5, and 0.8 as small, medium, and large, respectively.[4] For the ANOVA comparisons, we used the eta-square statistic (η^2 = sum of squares between groups/sum of squares total) and interpreted effect sizes of 0.01, 0.09, and 0.25 as small, medium, and large, respectively.[4]

Finally, a forward stepwise binary logistic regression analysis was conducted to determine which kinematic parameters could predict classification into either the injured or healthy group. Parameters identified to be significantly different between healthy and injured groups were considered for the regression model. Variables were excluded from the regression model if they were found to demonstrate differences between injured subgroups.

RESULTS

Injured Versus Healthy

The pooled data showed the injured runners to land with significantly more knee extension and ankle dorsiflexion (Table 3 and Figure 1). At midstance, the injured runners were found to have significantly greater forward trunk lean, CPD (Figure 2A), and hip adduction (Figures 2C and 3 and Table 4). Large effect sizes of 1.37, 0.89, and 0.87 were observed for CPD, hip adduction, and knee flexion at initial contact, respectively (Tables 3 and 4). Trunk forward lean at midstance and ankle dorsiflexion at initial contact demonstrated moderate effect sizes of 0.65 and 0.71, respectively (Tables 3 and 4). Chi-square tests found no significant difference in the distribution of foot strike patterns between the groups (P = .332). In the healthy group, there was a total of 17 forefoot and 19 rearfoot runners. In the injured group, there was a total of 27 forefoot and 45 rearfoot runners.

Injured Subgroups

The subgroup ANOVA was conducted to identify if there were differences between injured subgroups for variables identified as being different between the pooled injured and healthy groups. This analysis found no differences for ankle dorsiflexion and knee flexion at initial contact (Table 5). Furthermore, there were no differences in peak trunk forward lean and CPD at midstance (Table 5), indicating that these parameters were consistent across the injured subgroups. However, there was a significant difference between injured subgroups for peak hip adduction (Table 5). Post hoc least significant difference tests found the PFP (P = .018) and MTSS (P = .016) subgroups to have 3.1° and 3.2° more hip adduction than the ITBS group, respectively (Figure 2D).

Logistic Regression

The final variables identified as global kinematic contributors included knee flexion and ankle dorsiflexion at initial contact as well as trunk forward lean and CPD at midstance. All 4 variables were entered into the logistic regression model. The forward stepwise logistic regression model identified that CPD at midstance (odds ratio, 1.87 [95% CI, 1.41-2.49]; $P \leq .001$) and knee flexion at initial contact (odds ratio, 0.87 [95% CI, 0.78-0.97]; P = .012) were significant predictors of the classification as either healthy or injured, explaining 47% of the variance in the data (R^2 = 0.466). The most important predictor variable was CPD, with an 80% increase in the odds of being classified as injured for every 1° increase in pelvic drop. For knee flexion, there was a 23% reduction in the odds of being classified as injured for every 1° increase in knee flexion at initial contact.

DISCUSSION

This study identified a number of kinematic differences between the injured and healthy runners that were consistent across injured subgroups. In particular, the injured runners were found to demonstrate significantly greater peak CPD and forward trunk lean as well as a more extended knee and dorsiflexed ankle at initial contact (Tables 3-5 and Figures 1 and 3). We found CPD to be

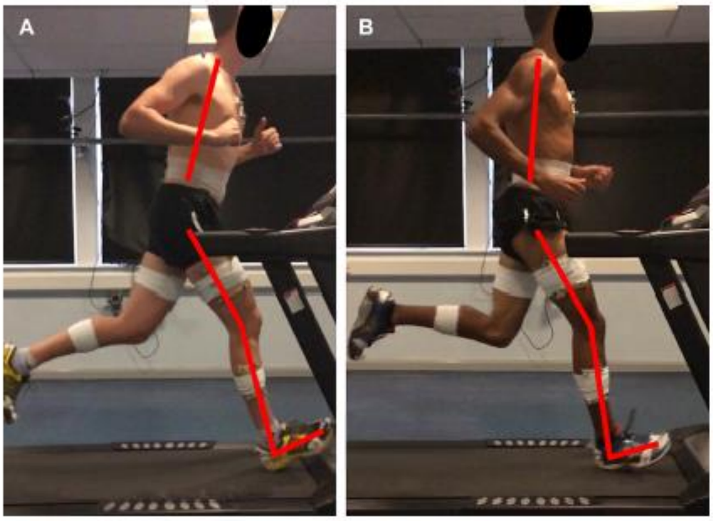

Figure 1. Two-dimensional representation of forward trunk lean, knee flexion, and ankle dorsiflexion at initial contact. (A) injured runner; (B) healthy runner.

TABLE 4
Peak Kinematic Angles at Midstance[a]

	Healthy	Injured	*P* Value	Effect Size
Trunk forward lean, deg	9.5 ± 2.9	12.0 ± 4.9	<.01[b]	0.65
Trunk ipsilateral lean, deg	3.6 ± 1.8	4.3 ± 2.6	.094	0.33
Pelvis anterior tilt, deg	5.0 ± 2.9	5.7 ± 3.8	.553	0.19
Contralateral pelvic drop, deg	3.7 ± 1.9	6.4 ± 2.1	<.01[b]	1.37
Hip adduction, deg	9.7 ± 3.5	13.0 ± 3.9	<.01[b]	0.89
Hip internal rotation, deg	4.4 ± 6.8	4.2 ± 8.0	.874	0.03
Knee flexion, deg	32.7 ± 4.9	32.3 ± 5.0	.556	0.09
Knee adduction, deg	−1.9 ± 3.1	−2.0 ± 3.5	.785	0.06
Knee external rotation, deg	6.7 ± 5.5	7.1 ± 6.9	.616	0.06
Ankle dorsiflexion, deg	22.3 ± 2.9	21.9 ± 4.3	.964	0.09
Rearfoot eversion, deg	2.6 ± 3.2	4.0 ± 3.5	.047	0.42

[a]Values are presented as mean ± SD unless otherwise specified. Angles represent the maximum joint angle between initial contact and toe-off.
[b]Statistical significance at $P < .01$.

the most important predictor variable when classifying runners as healthy or injured. These kinematic patterns may represent global kinematic contributors to soft tissue running injuries and together may define a pathological running gait.

Global Kinematic Contributors

Peak CPD was found to be the kinematic parameter most strongly associated with running injuries. Previous studies have associated CPD with PFP[52] and MTSS[26]; however, this study identified increased CPD among multiple different running-related injuries, including ITBS and AT (Figure 2B). Therefore, CPD may represent a global kinematic contributor and risk factor for many common soft tissue running injures.

It is likely that CPD will influence lower limb tissue stress at a number of different anatomic sites through a number of different mechanisms. For example, Tateuchi et al[43] identified that increasing CPD resulted in an increase in iliotibial band tension at the lateral femoral condyle. This will likely influence ITBS development

TABLE 5
Analysis of Variance Between Injured Subgroups[a]

	PFP	ITBS	MTSS	AT	*P* Value	Effect Size
Initial contact						
Knee flexion, deg	5.5 ± 4.6	6.6 ± 5.7	4.7 ± 5.2	7.4 ± 4.1	.365	0.05
Ankle dorsiflexion, deg	10.6 ± 3.9	7.1 ± 5.6	5.5 ± 9.2	5.6 ± 7.1	.088	0.09
Midstance						
Trunk forward lean, deg	11.9 ± 5.1	14.3 ± 5.5	10.9 ± 4.9	11.3 ± 3.4	.160	0.07
Contralateral pelvic drop, deg	6.4 ± 2.8	6.5 ± 2.4	6.6 ± 1.4	6.3 ± 1.9	.986	0.002
Hip adduction, deg	14.4 ± 4.5	11.3 ± 4.3	14.4 ± 1.6	12.2 ± 4.1	.032[b]	0.12

[a]Values are presented as mean ± SD unless otherwise specified. AT, Achilles tendinopathy; ITBS, iliotibial band syndrome; MTSS, medial tibial stress syndrome; PFP, patellofemoral pain.
[b]Statistical significance at $P < .05$.

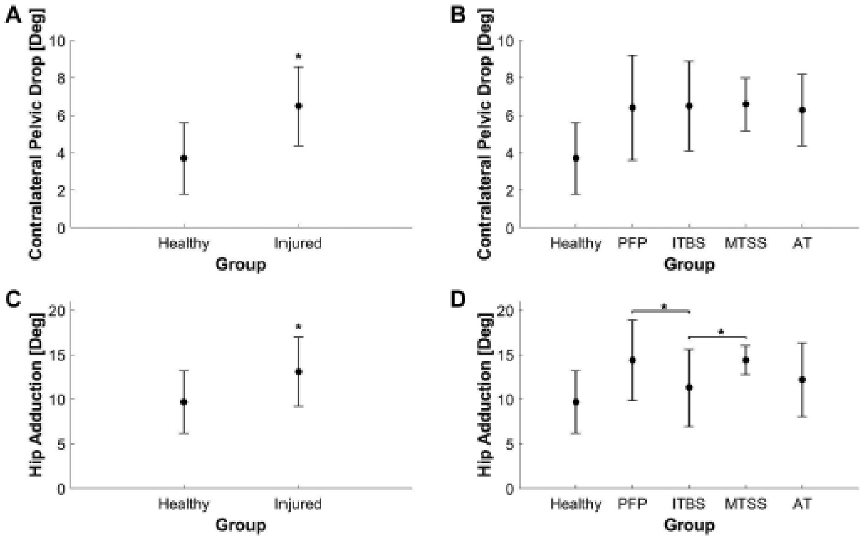

Figure 2. (A) Contralateral pelvic drop for healthy and injured groups. (B) Contralateral pelvic drop for healthy group and injured subgroups. (C) Hip adduction for healthy and injured groups. (D) Hip adduction for healthy group and injured subgroups. Whiskers represent ±1 SD. *Statistically significant differences for the (A, C) *t* test and (B, D) subgroup analysis of variance. The healthy group is shown in B and D for comparison purposes only. AT, Achilles tendinopathy; ITBS, iliotibial band syndrome; MTSS, medial tibial stress syndrome; PFP, patellofemoral pain.

through increased strain[19] and increased compression between the iliotibial band and lateral femoral condyle.[11] At the same time, an increase in iliotibial band tension will result in lateral displacement of the patella.[29] Lateral displacement of the patella will lead to a rise in patellofemoral joint stress, leading to PFP development,[36] while at the lower limb, increased CPD will result in a medial shift in the ground-reaction force relative to the knee joint center.[37,42] This may alter the force distribution through the lower limb, leading to increased bending forces on the medial tibia[5] and potentially alter the pressure distribution through the foot. This may contribute to the development of either MTSS or AT.[25,50]

One possible explanation for the increased CPD observed in the injured group could be reduced strength or neuromuscular function at the hip. Previous authors have reported the delayed onset of gluteus medius and maximus function in runners with PFP[51] and AT,[15] while others have reported reduced hip abductor strength in runners with ITBS,[16] PFP,[41] AT,[18] and MTSS.[48] The hip abductors, in particular, the gluteus medius, are thought to control frontal plane kinematics of the pelvis and

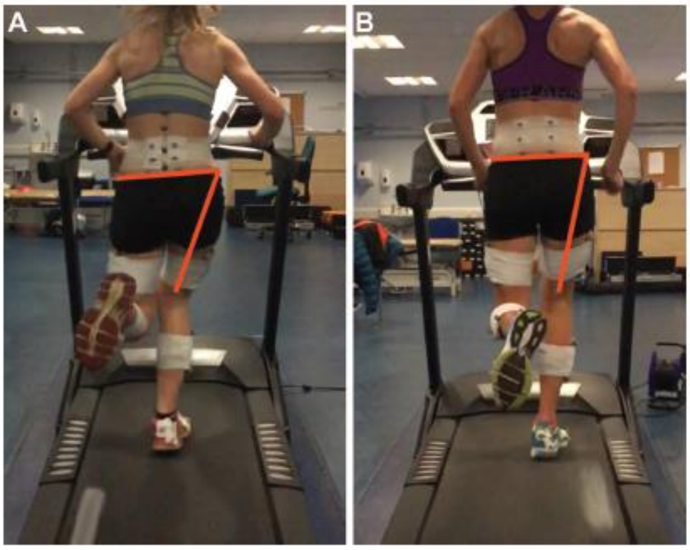

Figure 3. Two-dimensional representation of contralateral pelvic drop and hip adduction at midstance. (A) injured runner; (B) healthy runner.

hip.[40] Therefore, it is conceivable that reduced strength or neuromuscular function of the gluteus medius would lead to an inability to stabilize the pelvis in the frontal plane, causing increased CPD.

We also found the injured runners to land with greater knee extension and ankle dorsiflexion (Table 3 and Figure 1), which may influence tissue stress in a number of ways. First, in knee extension, the patella becomes vulnerable to lateral tilt and displacement, which may influence patellofemoral contact areas and joint stress during the early stance.[35] Second, an extended knee and dorsiflexed ankle at initial contact are typically associated with a greater distance between the center of mass and the foot at contact. A greater distance between the center of mass and foot, as well as larger ankle dorsiflexion angles, has been associated with increased knee joint loading and breaking impulse.[49] An extended knee at initial contact has also been reported to reduce the ability to attenuate impact forces during the early stance.[8] Collectively, it seems plausible that the extended lower limb posture at initial contact may influence impact loading and knee joint loading during the early stance.

One possible mechanism explaining the differences in forward trunk lean may be strength deficits around the gluteals and paraspinals. Previous studies have reported fatigue of the paraspinal and gluteal muscles to be associated with an increase in trunk forward lean during running[21] and drop landings.[23] Therefore, the reduced strength capacity of the gluteals and paraspinals may result in an inability to maintain an upright running posture among the injured runners.

Kinematic Subgroups

While hip adduction was found to be greater among the pooled injured group, the subgroup analysis revealed that this parameter differed across the injured subgroups (Table 5 and Figure 2, C and D). Specifically, we found hip adduction to be greater among subgroups of runners with PFP and MTSS compared with the ITBS subgroup (Figure 2D). This finding is in contrast to previous studies by Noehren et al[31] and Ferber et al,[12] who reported increased hip adduction among runners with ITBS. One potential reason for the contrasting findings may be the sex differences between studies. Hip adduction has been reported to be influenced by sex subgroups,[52] with greater hip adduction among female runners. In the current study, we included a mix of male and female participants, while Noehren et al[31] and Ferber et al[12] only included female participants. While we acknowledge that hip adduction may be an important kinematic risk factor for certain injuries, we believe that our data suggest that hip adduction may be more influential in specific subsets of runners and injuries rather than others.

Limitations

One limitation is that the study was retrospective, and therefore, it is not possible to conclude if the observed kinematic patterns are the cause of injury or the result of injury. Nevertheless, we ensured that all data were recorded before the onset of pain to minimize any possible effect of pain on the observed kinematic patterns. However, we cannot rule out the possibility that participants may have adapted their running kinematics in response to chronic injuries or in apprehension of the acute onset of pain. Therefore, we acknowledge that future prospective studies are required to further investigate whether the kinematic patterns observed within the current study are the cause or effect of injury. Another study limitation is the higher weekly mileage of the healthy group (Table 1). However, we believe that this could be considered a strength as previous research suggests that running greater than 40 miles per week is a risk factor for developing injuries.[46] On average, our healthy group was exceeding this threshold for more than 18 months before testing but remained injury free. Therefore, we believe that the healthy group may be representative of a healthy running gait to remain injury free at training loads exceeding the previously reported injury threshold. It is also important to note that this study was limited to a select number of common soft tissue running injuries, and therefore, these results may not apply to other injuries such as plantar heel pain, stress fractures, and muscle strains. Further research would be required to establish a link between the identified kinematic patterns and other running-related injuries.

Clinical Relevance

The findings from the present study may have a number of clinical implications. First, all of the identified kinematic parameters can be easily visualized using 2-dimensional (2D) gait analysis methods[9,10,34] (Figures 1 and 3). A number of recent publications have shown 2D assessments of CPD, hip adduction, trunk forward lean, and sagittal plane knee and ankle angles to be highly correlated with 3D measurement systems and to demonstrate high intratester and intertester reliability.[9,10,34] Therefore, it should be possible to use 2D measurement techniques to assess the biomechanical parameters that were associated with injuries in this study. Second, many of the identified global kinematic contributors to injuries can be modified through gait retraining. For example, CPD and hip adduction can be retrained using mirror feedback,[53] while knee and ankle angles are influenced by increasing cadence or modifying foot strike patterns.[30] Therefore, this study highlights a number of key kinematics that can be considered global contributors to running injuries and can be easily assessed and modified in clinical practice. This may assist clinicians in the development of rehabilitation programs for common running-related injuries.

CONCLUSION

This study identified a number of global kinematic contributors to common running injuries. In particular, we found injured runners to run with greater peak CPD and trunk forward lean as well as an extended knee and dorsiflexed ankle at initial contact. CPD appears to be the variable most strongly associated with common running-related injuries. The kinematic patterns identified as global contributors to injuries can be easily assessed and modified in clinical practice.

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16 Appendix G: Peer reviewed publication – A 10% Increase in Step rate Improves Running Kinematics and Clinical Outcomes in Runners with Patellofemoral Pain at 4 Weeks and 3 Months. *American Journal of Sports Medicine.*

A 10% Increase in Step Rate Improves Running Kinematics and Clinical Outcomes in Runners With Patellofemoral Pain at 4 Weeks and 3 Months

Christopher Bramah,*† MSc, MCSP, Stephen J. Preece,† PhD, Niamh Gill,† PhD, and Lee Herrington,† PhD, MCSP
Investigation performed at the School of Health Sciences, University of Salford, Salford, UK

Background: Aberrant frontal-plane hip and pelvis kinematics have been frequently observed in runners with patellofemoral pain (PFP). Gait retaining interventions have been shown to improve running kinematics and may therefore be beneficial in runners with PFP.

Purpose: To investigate whether a 10% increase in the running step rate influences frontal-plane kinematics of the hip and pelvis as well as clinical outcomes in runners with PFP.

Study Design: Case series; Level of evidence, 4.

Methods: Runners with PFP underwent a 3-dimensional gait analysis to confirm the presence of aberrant frontal-plane hip and/or pelvis kinematics at baseline. A total of 12 participants with frontal-plane hip and/or pelvis kinematics 1 standard deviation above a reference database were invited to undergo the gait retraining intervention. Running kinematics along with clinical outcomes of pain and functional outcomes were recorded at baseline, 4 weeks after retraining, and 3 months. Gait retraining consisted of a single session where step rate was increased by 10% using an audible metronome. Participants were asked to continue their normal running while self-monitoring their step rate using a global positioning system smartwatch and audible metronome.

Results: After gait retraining, significant improvements in running kinematics and clinical outcomes were observed at 4-week and 3-month follow-up. Repeated-measures analysis of variance with post hoc Bonferroni correction ($P < .016$) showed significant reductions in peak contralateral pelvic drop (mean difference [MD], 3.12° [95% CI, 1.88°-4.37°]), hip adduction (MD, 3.99° [95% CI, 2.01°-5.96°]), and knee flexion (MD, 4.09° [95% CI, 0.04°-8.15°]) as well as significant increases in self-reported weekly running volume (MD, 13.78 km [95% CI, 4.62-22.93 km]) and longest run pain-free (MD, 6.84 km [95% CI, 3.05-10.62 km]). Friedman test with a post hoc Wilcoxon signed-rank test showed significant improvements on a numerical rating scale for worst pain in the past week and the Lower Extremity Functional Scale.

Conclusion: A single session of gait retraining using a 10% increase in step rate resulted in significant improvements in running kinematics, pain, and function in runners with PFP. These improvements were maintained at 3-month follow-up. It is important to assess for aberrant running kinematics at baseline to ensure that gait interventions are targeted appropriately.

Registration: NCT03067545 (ClinicalTrials.gov identifier)

Keywords: patellofemoral pain; running; gait retraining; kinematics; step rate

*Address correspondence to Christopher Bramah, MSc, MCSP, School of Health Sciences, University of Salford, Blatchford Building, Salford, M6 6PU, UK (email: c.a.bramah@salford.ac.uk) (Twitter: @chrisbramah).

†School of Health Sciences, University of Salford, Salford, UK.

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Recreational running is an increasingly popular method of physical activity, with participation rates growing annually. Although running offers several health benefits, it also poses a considerable risk of injury to the musculoskeletal system. Overall injury incidence rates are reported to range between 19% and 78% among recreational runners,[37] with recurrence rates in 20% to 70% of all cases.[38] Of all running injuries, patellofemoral pain (PFP) is considered the most common running-related knee injury,[36] with incidence and prevalence rates as high as 20.8% and 22.7%, respectively.[34]

PFP is known to have a multifactorial cause, with aberrant running mechanics identified as one risk factor.[21,22,24]

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Runners with PFP have been reported to demonstrate increased hip adduction (HADD),[24,25,41] hip internal rotation,[25] and contralateral pelvic drop (CPD)[41] when compared with injury-free controls. It is thought that aberrant frontal- and transverse-plane running kinematics may result in lateral tracking of the patella, leading to a rise in patellofemoral joint stress.[30] When exposed to repeat loading cycles during running, the knee may experience damage to the underlying chondral surface, stress within the subchondral bone, and excitation of nociceptors, leading to pain and injury.[2]

Gait retraining is a clinical intervention that targets aberrant running kinematics within the rehabilitation of PFP. Current evidence has shown improvements in kinematics and clinical outcomes after mirror retraining,[42] real-time feedback,[26] and transitioning to a forefoot contact.[32,33] However, the use of mirror retraining and real-time feedback is restricted to clinical and laboratory settings, limiting their practical applicability, while transitioning to a forefoot strike has been shown to increase Achilles tendon and ankle joint loading, which may increase the risk of lower limb injury.[1] Furthermore, these studies often utilize a faded feedback design consisting of 8 sessions over a 2-week period, requiring close clinical supervision. Therefore, there is a need for gait retraining methods that can be easily integrated outside of a laboratory setting while providing positive clinical and biomechanical outcomes.

Increasing step rate may be one method of gait retraining that could be integrated outside of a laboratory setting. Through the use of global positioning system (GPS) smartwatches and mobile metronome applications, runners may be able to self-retrain and monitor their step rate without the need for close clinical supervision.[19,40] Currently, only 3 studies have investigated the effects of increasing the step rate among runners with PFP.[11,12,20] Neal et al[20] reported improved frontal-plane hip and pelvis kinematics along with reductions in pain but did not investigate whether improvements were maintained beyond the 6-week follow-up period. Esculier et al[12] reported gait retraining to be no more effective than education on load management, and Dos Santos et al[11] reported minimal changes in pain after a 2-week retraining period. Furthermore, both Esculier et al and Dos Santos et al did not report any change in frontal-plane hip and pelvis kinematics after the retraining period. Therefore, questions remain regarding the clinical effectiveness of increasing the step rate and whether step rate retraining results in long-term kinematic adaptations among runners with PFP.

The aim of the current study was to investigate whether a 10% increase in the running step rate influences frontal-plane kinematics of the hip and pelvis as well as clinical outcomes in runners with PFP. Rather than use a clinically monitored faded feedback design, we sought to investigate whether runners can self-administer gait retraining interventions using an audible metronome and a GPS smartwatch. We hypothesized that a 10% increase in step rate would result in significant reductions in frontal-plane hip and pelvis kinematics and improvements in clinical outcomes and that these changes would be present at short- and long-term follow-up.

METHODS

Participants

Participants were recruited through advertisements at local sports injury clinics, running clubs, and a university-based running clinic. Ethical approval for the study was obtained via the local ethics committee, and all participants provided written informed consent before participation. This study was registered as a clinical trial (ClinicalTrials.gov registration No. NCT03067545) with enrollment for the trial between March 2017 and December 2018. An a priori sample size calculation was conducted using data from a previous gait retraining study identifying a 2.3° reduction in CPD after retraining with an effect size of 1.09.[26] Using G*Power software, we calculated that 12 participants would be required to detect an effect size of 1.25 with a power of 0.8 and an adjusted critical alpha of .016. This calculation was based on the use of paired tests to detect differences in peak CPD, which were similar to changes observed in previous studies after gait retraining (2.3°)[26] and also of similar magnitude to differences observed between injured and healthy runners (2.7°).[4]

Inclusion/Exclusion Criteria

All participants were required to own a GPS smartwatch or running watch capable of monitoring step rate. Participants were included in the gait retraining intervention based on a 3-stage assessment process. First, a subjective assessment and clinical examination were used to confirm the presence of PFP. Once the diagnosis of PFP was confirmed, a 3-dimensional (3D) gait analysis was conducted to confirm the presence of aberrant hip and/or pelvis kinematics. To ensure that the injury diagnosis met the consensus definition of a running-related injury,[43] participants had to report an insidious onset of anterior knee pain during running lasting for a minimum of 3 months, causing a self-reported restriction to either their running volume or duration. Participants were required to be running a minimum of twice per week, with their worst pain rated a minimum of 3 out of 10 on a numerical rating scale (NRS) for pain (0 = no pain, 10 = worst possible pain). Pain must also have been reproduced by ≥1 of the additional activities of squatting, kneeling, prolonged sitting, or ascending or descending stairs. Participants were excluded if they reported having any known medical condition, had undergone prior musculoskeletal surgery, had a neurological impairment, had diagnosed knee osteoarthritis, had a structural deformity of the knee, had an onset of knee pain caused by trauma or any other sporting activity, had ceased running, or were undergoing additional treatment outside of the study. To control for training errors as a potential underlying cause of injury, participants were also excluded if they reported the onset of symptoms to occur after an increase in their weekly training volume ≥30%.[23]

After the subjective assessment, participants were invited to undergo a clinical examination led by the lead clinician (C.B.) to confirm the diagnosis of PFP in accordance with previously published diagnostic criteria.[9] Specifically, for inclusion in the study, pain must have been retropatellar or peripatellar in nature and reproduced on squatting with the exclusion of any patellar instability, ligamentous or meniscal injury.[9] Pain on squatting has been shown to have a sensitivity of 91% and a negative predictive value of 74%, suggesting this examination to be the best available test for PFP.[9,27] A combination of additional, but nonessential, clinical tests was used to further increase the diagnostic accuracy of PFP.[8] Tests included patellar compression, patellar apprehension, pain on palpation of the lateral patellar facet, and pain on resisted quadriceps contraction in 30° of knee flexion.[8,27] These tests are known to have low sensitivity and specificity when used in isolation and were not used as a sole diagnostic criterion for PFP.[8,27]

Once the diagnosis of PFP was confirmed, each participant underwent an initial 3D gait analysis as outlined below and completed clinical outcome measures to monitor pain and functional improvements. Clinical outcome measures included the Lower Extremity Functional Scale (LEFS), previously validated for use in PFP,[3] as well as self-reported worst pain experienced in the past week using the NRS (0 = no pain, 10 = worst possible pain). Additional outcomes monitored were self-reported longest distance run pain-free and total weekly running volume.

Kinematic Data Collection

Kinematic data were collected from all participants with confirmed PFP while running on a treadmill (F63; Sole Fitness) at 3.2 m/s wearing their own running shoes. After a 5-minute warm-up period, 30 seconds of kinematic data were collected using a 12-camera Oqus system (240 Hz; Qualisys). A total of 9 anatomic segments were tracked following a previously published protocol.[17,31] Segments included the thorax; pelvis; and bilateral thigh, shank, and foot segments. Further details of the markers used to track each segment and the precise definition of the anatomic coordinate systems are provided in previous publications.[13,17,31]

Raw kinematic data were low pass filtered at 10 Hz. Intersegmental kinematics, along with motions of the pelvis and thorax with respect to the laboratory system, were calculated using a 6 degrees of freedom model with Visual3D software (C-Motion). Gait events were defined using a kinematic approach in which initial contact was defined as the first vertical acceleration peak of either the heel or metatarsal markers and toe-off defined as the vertical jerk peak of the second metatarsal marker.[14] Gait events were subsequently used to segment each kinematic signal into a minimum of 10 consecutive gait cycles. An ensemble average for each signal was created and selected kinematic parameters derived from the ensemble average curves. This latter processing was carried out using a custom Matlab script (MathWorks).

Participants were invited to participate in the gait retraining study providing that they demonstrated aberrant hip and/or pelvis kinematics during the initial 3D gait analysis. Aberrant hip and pelvis kinematics were defined as peak HADD and CPD angles $\geq$1 standard deviation above the mean of a previously published database of healthy recreational runners running at the same speed[4] (qualifying criteria = CPD $\geq$5.6° and/or HADD $\geq$13.2°). Runners who did not meet the kinematic inclusion criteria were not included in the study and were referred to a health care professional for further management.

Retraining Protocol

All participants included within the retraining protocol completed a single 10-minute retraining session conducted immediately after the initial 3D gait analysis. During the retraining session, participants were asked to run at the same speed with a 10% increase in their original step rate. Step rate was calculated as the number of foot contacts per minute. During the first 5 minutes of the retraining protocol, participants were instructed to match their footsteps to an audible metronome set to the new step rate. For the final 5 minutes of the retraining session, the audible metronome was removed, and participants were instructed to continue running at the increased step rate. Throughout this time, participants were monitored by the lead researcher to ensure that they were able to maintain the higher step rate and the metronome reintroduced if they failed to do so.

After the retraining session, participants were provided with instructions for self-administration and monitoring of the increased step rate. Specifically, during the first 2 weeks, participants were instructed to continue using a free downloadable metronome application set to the new step rate. During the third and fourth weeks, participants were instructed to continue running without the use of the metronome but were instructed to self-monitor their cadence using their GPS smartwatch. Participants were permitted to increase their running volume at any point in the retraining period provided that any knee pain experienced was rated below 3 of 10 on an NRS.

All participants were invited to follow-up 3D gait analysis sessions at 4 weeks and 3 months after the initial assessment. This follow-up period allowed us to investigate whether kinematic changes could be maintained across a time frame comparable with previous gait retraining studies.[42] The follow-up sessions were completed following the same kinematic testing procedures as the first visit, recording the same clinical outcome measurements. After the 4-week follow-up assessment, participants were instructed to continue running without the use of the metronome. No restrictions to training parameters were provided; participants were instead instructed to increase their training volume or pace and change surfaces as they saw fit provided that any pain experienced was rated lower than 3 of 10 on the NRS.

Data Analysis

Several kinematic parameters were selected for data analysis. Kinematic parameters included peak CPD, HADD, hip internal rotation, and knee flexion. These parameters were selected based on previous research highlighting associations with these parameters and PFP.[4,20,25,41] Peak angles at midstance were defined as the maximum joint angles between initial contact and toe-off. Stride rate was also included within the analysis, measured as steps per minute, along with clinical outcome measures of worst pain experienced in the past week using the NRS, longest distance run pain-free, total weekly running volume, and LEFS score.

Statistical Analysis

Repeated-measures analysis of variance was used to assess for differences in kinematic parameters between the initial assessment (baseline), 4-week follow-up, and 3-month follow-up, with a critical alpha of <.05. When significant differences were observed, post hoc Bonferroni testing, with an adjusted critical alpha of <.016, was used to identify differences between time points. Clinical outcomes of the NRS and LEFS were analyzed using the Friedman test for nonparametric data with a post hoc Wilcoxon signed-rank test. Effect sizes were calculated for pairwise comparisons using the Cohen *d* and interpreted as 0.2, 0.5, and 0.8 as small, medium, and large, respectively.[6]

RESULTS

A total of 33 participants met the initial subjective inclusion criteria and were invited for a clinical examination. After the clinical examination, 18 were diagnosed as having PFP and invited to take part in the 3D gait analysis. After the 3D gait analysis, 12 participants met the inclusion criteria and were enrolled in the gait retraining study. There were 2 participants who dropped out of the study between the 4-week and 3-month follow-up points. The first participant failed to attend the 3-month follow-up and did not respond to contact, and the second developed a tibial stress fracture on the same limb and was unable to continue the study. Both participants were included in the final analysis using a last observation carried forward method[35] (Table 1).

TABLE 1
Participant Characteristics[a]

	Value
Male/female sex, n	4/8
Age, y	39.92 ± 6.50
Weight, kg	61.03 ± 6.48
Height, cm	170.33 ± 6.98
Usual weekly running volume, km	29.03 ± 8.11

[a]Values are presented as mean ± SD unless otherwise specified.

Kinematics

Repeated-measures analysis of variance showed a significant effect of time for several kinematic parameters (Table 2). In particular, there were significant increases in step rate and reductions in peak CPD, HADD, and knee flexion after the step rate intervention (Table 2). The post hoc test revealed that step rate significantly increased by an average of 11.2% at 4 weeks (mean difference [MD], 18.6 steps/min [95% CI, 11.97-25.23 steps/min]) and 9.2% at 3 months (MD, 15.1 steps/min [95% CI, 10.64-19.57 steps/min]) when compared with baseline. There was a significant 3.12° reduction in CPD (MD, 3.12° [95% CI, 1.88°-4.37°]) and 3.99° reduction in HADD (MD, 3.99° [95% CI, 2.01°-5.96°]) at 4 week follow-up, which was also significant at 3 months when compared with baseline for both CPD (MD, 2.7° [95% CI, 1.4°-4.1°]) and HADD (MD, 2.8° [95% CI, 0.4°-5.4°]). Similarly, there was a significant reduction in peak knee flexion during stance phase at 4 weeks (MD, 4.10° [95% CI, 0.04°-8.15°]) and 3 months (MD, 4.15° [95% CI, 0.81°-7.48°]). No significant differences were observed between the 4-week and 3-month follow-up time points for any of the kinematic parameters (Table 2).

Clinical and Functional Outcomes

All clinical and functional outcomes demonstrated statistically significant improvements. Specifically, there was a significant reduction in pain scores on the NRS from an average of 6.2 out of 10 at baseline to 1.0 and 0.3 at 4 weeks and 3 months, respectively (χ^2 = 21.38; P < .01) (Figure 1), which is above the minimal clinically important difference of 1.2 points.[28] The LEFS demonstrated a statistically significant improvement from 62.3 at baseline to 76.6 at 4 weeks and 79.7 at 3 months (χ^2 = 22.29; $P \leq$ .01) (Figure 1). When compared with baseline, this was a 14.3- and 17.4-point improvement at 4 weeks and 3 months, respectively, which is above the minimal clinically important difference of 9 points.[3] All participants demonstrated a significant increase in total weekly running volume (MD, 13.78 km [95% CI, 4.62-22.93 km]) and longest distance run pain-free (MD, 6.84 km [95% CI, 3.05-10.62 km]) from baseline to 4-week and 3-month follow-up (Table 3).

DISCUSSION

The aim of this study was to investigate whether a 10% increase in the step rate would improve kinematics and clinical outcomes among runners with PFP who demonstrate aberrant frontal-plane pelvis and hip kinematics. In support of our hypothesis, we observed significant reductions in frontal-plane pelvis and hip kinematics as well as significant reductions in pain and improvements in function and running at 4 weeks, which appeared to be maintained at 3-month follow-up.

After the step rate increase, we observed a 3.12° and 3.99° reduction in CPD and HADD, respectively (Table 2), which may offer a mechanical explanation for the

TABLE 2
Kinematic Parameters[a]

	Baseline	4 wk	3 mo	*P* Value (ANOVA)	Pairwise Effect Size (Cohen *d*)	
Stride rate, steps/min	165.93 ± 7.38	184.53 ± 10.10[b]	181.04 ± 7.78[b]	<.01	Baseline to 4 wk	2.10
					Baseline to 3 mo	1.99
					4 wk to 3 mo	0.38
Peak CPD, deg	7.46 ± 1.81	4.34 ± 2.47[b]	4.73 ± 2.95[b]	<.01	Baseline to 4 wk	1.44
					Baseline to 3 mo	1.11
					4 wk to 3 mo	0.14
Peak HADD, deg	15.95 ± 2.75	11.96 ± 1.77[b]	13.09 ± 3.20[b]	<.01	Baseline to 4 wk	1.72
					Baseline to 3 mo	0.95
					4 wk to 3 mo	0.43
Peak hip internal rotation, deg	4.06 ± 8.17	4.18 ± 9.35	4.42 ± 7.89	.93	Baseline to 4 wk	0.01
					Baseline to 3 mo	0.04
					4 wk to 3 mo	0.02
Peak knee flexion, deg	33.74 ± 5.25	29.64 ± 3.23[b]	29.59 ± 3.16[b]	<.01	Baseline to 4 wk	0.94
					Baseline to 3 mo	0.96
					4 wk to 3 mo	0.02

[a]Values are presented as mean ± SD unless otherwise specified. Pairwise effect sizes are interpreted as 0.2 = small, 0.5 = medium, and 0.8 = large. ANOVA, analysis of variance; CPD, contralateral pelvic drop; HADD, hip adduction.
[b]Significant difference when compared with baseline at $P < .016$ after Bonferroni correction.

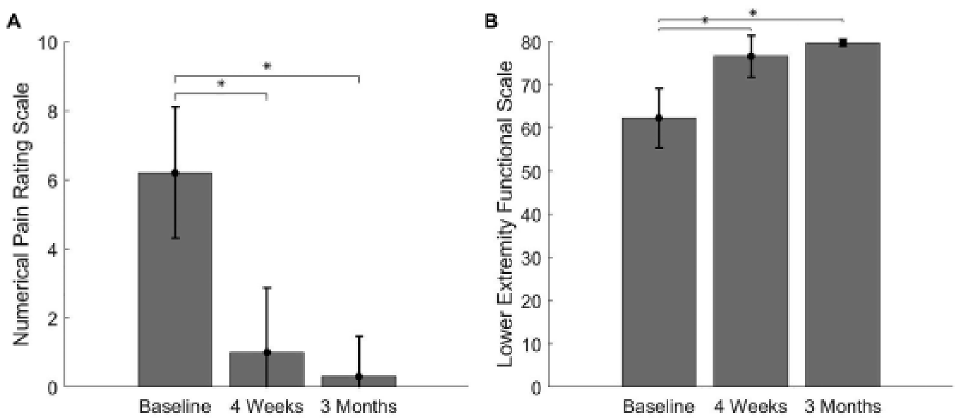

Figure 1. Clinical outcomes at baseline, 4 weeks, and 3 months. *Statistically significant difference when compared with baseline at $P < .016$. Error bars represent ±1 standard deviation.

improved clinical outcomes seen in this study. These changes are greater than those observed in previous step rate studies,[11,20] with this being the first study to highlight that kinematic adaptations are maintained at longer term follow-up. CPD and HADD have been cited as kinematic risk factors for PFP.[4,21,24,25,41] It is thought that CPD will give rise to an increase in iliotibial band tension, resulting in lateral displacement of the patella,[15,18] while HADD would cause the femur to shift medially under the patella.[29] This would result in elevated contact pressure between the patella and lateral facet, leading to elevated joint stress and potential injury.[30] Therefore, it is possible that the reductions in CPD and HADD after an increase in the step rate would contribute to reduced lateral displacement of the patella and a corresponding reduction in patellofemoral joint stress.

Similarly, the reduction in peak knee flexion during stance phase may also contribute to improvements in clinical outcomes. Peak knee flexion during stance phase has been shown to influence patellofemoral joint reaction force, explaining up to 64% of the variance in peak patellofemoral joint loading.[16] Smaller knee flexion angles at midstance will likely reduce the external joint force as well as reduce the demand on the surrounding musculature.[16] In the current study, we observed a 4.1° reduction in peak knee flexion (Table 2). Given the work of Lenhart et al,[16]

TABLE 3
Functional Outcomes[a]

Parameter	Baseline	4 wk	3 mo	P Value	Pairwise Effect Size (Cohen d)	
Total distance per week, km	13.34 ± 9.83	27.11 ± 11.25[b]	28.33 ± 12.96[b]	<.01	Baseline to 4 wk	1.30
					Baseline to 3 mo	1.30
					4 wk to 3 mo	0.10
Longest run pain-free, km	2.03 ± 1.19	8.87 ± 4.39[b]	11.33 ± 6.42[b]	<.01	Baseline to 4 wk	2.10
					Baseline to 3 mo	2.01
					4 wk to 3 mo	0.44

[a]Values are presented as mean ± SD unless otherwise specified. Pairwise effect sizes are interpreted as 0.2 = small, 0.5 = medium, and 0.8 = large.
[b]Statistically significant difference when compared with baseline at $P < .016$.

this magnitude of change is likely to contribute to reductions in peak patellofemoral joint force. These reductions, combined with the reductions in peak HADD and CPD, will likely lead to significant reductions in patellofemoral joint stress, which may explain the observed improvements in clinical outcomes within the present study.

We suggest that the improved frontal-plane hip and pelvis kinematics may be explained by alterations in neuromuscular activity of the hip. Willson et al[39] found that runners with PFP demonstrate significantly delayed onset of the gluteus medius, which had a moderate correlation with HADD excursion. It is hypothesized that delayed muscle activation of the gluteus medius during the stance phase of running would result in a loss of neuromuscular stiffness about the hip and pelvis, leading to a loss of frontal-plane stability.[39] Increasing step rate by 10% has been shown to directly influence the preactivation of gluteal muscles.[5] Specifically, Chumanov et al[5] reported significantly increased gluteus medius and maximus muscle activity in late swing phase, just before initial foot contact after a 10% increase in step rate. Considering the role that the gluteus medius plays in frontal-plane stability of the hip and pelvis, it is likely that the earlier onset of the gluteal muscles would result in increased neuromuscular stability during the stance phase of gait. This would likely explain the mechanical improvements of reduced CPD and HADD observed in the present study.

Reductions in peak knee flexion may also be explained by alterations in neuromuscular activity at the knee. Increasing the step rate has been shown to result in greater preactivation of the hamstrings, vastus lateralis, and rectus femoris muscles during late swing phase.[5] It is thought that these changes in neuromuscular coordination contribute to a more extended knee throughout the stance phase, reducing peak knee flexion angles.[5,16]

In contrast to previous studies, we did not identify differences in peak hip internal rotation after gait retraining. Neal et al[20] reported a 5.1° reduction in peak hip internal rotation after a 10% increase in step rate, whereas in the present study, we did not observe more than a 0.5° change. This may be explained by our baseline inclusion criteria of increased HADD and CPD rather than hip internal rotation. Participants within this study demonstrated 4.06° of hip internal rotation at baseline, which is less than the 9.1° reported in the study by Neal et al[20] and similar to the 4.4° reported in the database of healthy runners used for baseline reference values.[4] Therefore, it is possible that participants in the present study did not demonstrate increased hip internal rotation angles at baseline and thus would be unlikely to demonstrate any change.

An interesting finding was the magnitude of clinical improvements made by participants. Specifically, participants reported their worst pain to be, on average, 1.0 out of 10 at 4-week follow-up and 0.3 out of 10 at 3-month follow-up (Figure 1). This is greater than the minimal clinically important difference of 1.2 points[28] and greater than improvements seen in previous step rate studies, which have reported average NRS scores of 3,[11] 3.8,[12] and 2.9[20] out of 10 after retraining. We also observed significant improvements in function, with all runners reporting an increase in their weekly running volume and longest distance run pain-free as early as 4 weeks (Table 3), as well as a 17.4-point improvement on the LEFS at 3 months, exceeding the minimal clinically important difference of 9 points.[3] This contrasts with previous step rate studies, with 1 study reporting participants to be running less than their preinjury status at 20-week follow-up[12] and another study reporting less than a 9-point improvement on the LEFS.[11]

The reason for the magnitude of kinematic and clinical improvements in the present study compared with previous step rate studies may be because of the differences in inclusion criteria. In the present study, we specifically targeted participants who demonstrated aberrant kinematics at baseline. We did this to account for the multifactorial etiology of PFP and ensure that the appropriate underlying injury driver was targeted through the gait intervention. Failure to consider alternative causes of injury would likely result in the inclusion of biomechanical nonresponders in the retraining group. As such, these participants would be unlikely to demonstrate significant clinical improvements. Willy et al[42] and Noehren et al[26] are the only previous studies to use a similar inclusion criterion, with their results showing a similar magnitude of clinical improvement. Therefore, we suggest that future research should aim to establish the underlying pathological driver to appropriately target clinical interventions and that this be mirrored in clinical practice.

Clinical Relevance

In contrast to previous gait retraining studies, we opted to allow runners to self-administer and self-monitor their retraining using a metronome application and feedback from a GPS smartwatch. This proved successful, as all runners were able to maintain an increased cadence at 4-week and 3-month follow-up. Furthermore, at 4-week follow-up, all participants reported that they did not use the metronome beyond the first week and instead self-monitored their cadence using their GPS smartwatch. Previous studies have utilized a faded feedback design in which feedback is gradually removed over 8 sessions across a 2-week period. Although faded feedback designs have proven clinically effective, they require close clinical supervision and are restricted to clinical and laboratory settings. The present study demonstrates that simple step rate retraining can be applied outside of the laboratory and with minimal clinical contact. Importantly, 2-dimensional measures of CPD and HADD have been shown to be valid and reliable when compared with 3D measurements.[10] Therefore, assessment of running kinematics and gait retraining can be easily integrated into clinical practice and a participant's normal running routine.

Limitations

One limitation of the present study is the lack of a control group, making it difficult to ascertain whether the observed findings are true intervention effects. However, participants in the present study had an average NRS pain score of 6.2 out of 10 at baseline and a minimum symptom duration of 3 months, which has been reported to be predictive of poor prognosis at long-term follow-up.[7] Therefore, it is unlikely that participants would have experienced the magnitude of symptom improvement without a clinical intervention. Furthermore, the kinematic differences observed between baseline and the 4-week and 3-month time points were statistically significant, with large effect sizes above the standard error of measurement previously reported for these parameters.[17] These differences were of a similar magnitude to those observed previously between injured and healthy runners (2.7°)[4] and after gait retraining interventions (2.3°).[26] Therefore, we believe that the differences observed represent true intervention effects and could be considered biomechanically important.

A second limitation is the small participant numbers, which makes the statistical interpretation of small differences difficult. We observed no significant kinematic changes between 4-week and 3-month follow-up. We acknowledge that the small sample size limited the statistical power to detect small differences between these time points. However, the magnitude of the differences in biomechanical outcomes between these 2 follow-up points was typically small (Table 2) and lower than the standard error of measurement previously reported for these parameters.[17] Therefore, we believe that subtle differences between these time points are unlikely to be clinically important. However, we acknowledge that future randomized controlled trials with larger participant numbers are necessary to further validate our findings and confirm that kinematic changes are maintained over longer time periods.

It is important to note that 1 participant dropped out of the intervention after suffering a tibial stress fracture. This participant reported that the injury onset occurred after a sudden increase in training volume in preparation for a half-marathon. As we did not control participant progression of training volume, it is possible that the injury could be the result of training behaviors rather than a response to the intervention. As such, we recommend that future clinical interventions provide participants with specific advice on the safe progression of running volume to reduce the risk of further injuries.

CONCLUSION

The results of this study highlight that a 10% increase in step rate improves running kinematics and clinical outcomes at 4 weeks, which are maintained at 3 months, among runners with PFP. Therefore, step rate retraining appears to be a clinically effective intervention in the rehabilitation of PFP and can easily be integrated into clinical practice. It is important to assess running kinematics at baseline to ensure that interventions are appropriately targeted.

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17 Appendix H: Lower Extremity Functional Scale

EMORY HEALTHCARE
EMORY PHYSICAL THERAPY

Lower Extremity Functional Scale (LEFS)

Source: Binkley JM, Stratford PW, Lott SA, Riddle DL. The Lower Extremity Functional Scale (LEFS): scale development, measurement properties, and clinical application. North American Orthopaedic Rehabilitation Research Network. *Phys Ther*. 1999 Apr;79(4):371-83.

The Lower Extremity Functional Scale (LEFS) is a questionnaire containing 20 questions about a person's ability to perform everyday tasks. The LEFS can be used by clinicians as a measure of patients' initial function, ongoing progress and outcome, as well as to set functional goals.

The LEFS can be used to evaluate the functional impairment of a patient with a disorder of one or both lower extremities. It can be used to monitor the patient over time and to evaluate the effectiveness of an intervention.

Scoring instructions

The columns on the scale are summed to get a total score. The maximum score is 80.

Interpretation of scores

- The lower the score the greater the disability.
- The minimal detectable change is 9 scale points.
- The minimal clinically important difference is 9 scale points.
- % of maximal function = (LEFS score) / 80 * 100

Performance:

- The potential error at a given point in time was +/- 5.3 scale points.
- Test-retest reliability was 0.94.
- Construct reliability was determined by comparison with the SF-36. The scale was found to be reliable with a sensitivity to change superior to the SF-36.

Instructions

We are interested in knowing whether you are having any difficulty at all with the activities listed below **because of your lower limb problem** for which you are currently seeking attention. Please provide an answer for **each** activity.

Today, *do you* or *would you* have any difficulty at all with:

Activities	Extreme difficulty or unable to perform activity	Quite a bit of difficulty	Moderate difficulty	A little bit of difficulty	No difficulty
1. Any of your usual work, housework or school activities.	0	1	2	3	4
2. Your usual hobbies, recreational or sporting activities.	0	1	2	3	4
3. Getting into or out of the bath.	0	1	2	3	4
4. Walking between rooms.	0	1	2	3	4
5. Putting on your shoes or socks.	0	1	2	3	4
6. Squatting.	0	1	2	3	4
7. Lifting an object, like a bag of groceries from the floor.	0	1	2	3	4
8. Performing light activities around your home.	0	1	2	3	4
9. Performing heavy activities around your home.	0	1	2	3	4
10. Getting into or out of a car.	0	1	2	3	4
11. Walking 2 blocks.	0	1	2	3	4
12. Walking a mile.	0	1	2	3	4
13. Going up or down 10 stairs (about 1 flight of stairs).	0	1	2	3	4
14. Standing for 1 hour.	0	1	2	3	4
15. Sitting for 1 hour.	0	1	2	3	4
16. Running on even ground.	0	1	2	3	4
17. Running on uneven ground.	0	1	2	3	4
18. Making sharp turns while running fast.	0	1	2	3	4
19. Hopping.	0	1	2	3	4
20. Rolling over in bed.	0	1	2	3	4
Column Totals:	0	1	2	3	4