PH.D THESIS

# The Biomechanical Evaluation of Athletic Groin Pain

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DCU

## Declaration

I hereby certify that this material, which I now submit for assessment on the programme of study leading to the award of Doctor of Philosophy is entirely my own work, that I have exercised reasonable care to ensure that the work is original, and does not to the best of my knowledge breach any law of copyright, and has not been taken from the work of others save and to the extent that such work has been cited and acknowledged within the text of my work.

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Shane Gore

June 11, 2018

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# Abstract

**Introduction:** Athletic groin pain (AGP) is a common injury in sports involving repetitive twisting, kicking and turning (Werner et al., 2009; Thorborg et al., 2017). In male soccer for example, AGP incidence accounts for 4 -19% of all injuries (Waldén, Hägglund and Ekstrand, 2015). Despite this, there remains a dearth of literature investigating the three-dimensional biomechanics of AGP. The primary aim of this PhD thesis is to evaluate the biomechanical factors affected by AGP to enhance our understanding of this injury.

**Methods:** This thesis incorporates the work from eight investigations including one published systematic review incorporated in the review of literature, two methodological studies and five experimental investigations. As a body of work, this is the one of the largest investigations of AGP biomechanics conducted to date as it includes an investigation of over 200 AGP patients and 85 uninjured subjects.

**Results and Conclusion:** Multiple biomechanical factors were identified that may be related to AGP. The mechanics of the trunk and hip are commonly affected by this injury, particularly in the frontal plane, and may represent targets for rehabilitation. The ankle joint was also consistently and strongly identified as being affected by AGP but generally failed to change significantly from pre- to post- rehabilitation. As such, future research is warranted to ascertain if AGP rehabilitation could be further enhanced with a focus on improving ankle function. Given that variability and complexity appear to be affected by AGP with large effect sizes, it is suggested that the rehabilitation of AGP should follow a dynamic systems theory approach (Newell 1986).

The results from this PhD thesis have made novel contributions on an empirical, theoretical, methodological and practical level and the findings can help guide clinical practice and the design of future prospective research.

# List of Publications

## Journal articles

Gore, S.J., Franklyn-Miller, A., Richter, C., Falvey, E.C. King, E., and Moran, K.A., 2018. Is stiffness related to athletic groin pain ? Scandinavian Journal of Medicine & Science in Sports, In press.

Baida, S., Gore, S., Franklyn-Miller, A., Moran, K., 2017. Does the amount of lower extremity movement variability differ between injured and uninjured populations? A systematic review. Scandinavian Journal of Medicine & Science in Sports, In press.

Franklyn-Miller, A., Richter, C., King, E., Gore, S., Moran, K., Strike, S. and Falvey, E.C., 2017. Athletic groin pain (part 2): a prospective cohort study on the biomechanical evaluation of change of direction identifies three clusters of movement patterns.British journal of sports medicine,51(5), pp.460-468.

Gore, S.J., Marshall, B.M., Franklyn-Miller, A.D., Falvey, E.C. and Moran, K.A., 2016. The Number of Trials Required to Obtain a Representative Movement Pattern During a Hurdle Hop Exercise. Journal of applied biomechanics,32(3), pp.295-300.

Marshall, B., Franklyn-Miller, A., Moran, K., King, E., Richter, C., Gore, S., Strike, S. and Falvey, E.C, 2015. Biomechanical symmetry in elite rugby union players during dynamic tasks: an investigation using discrete and continuous data analysis techniques. BMC sports science, medicine and rehabilitation,7(1), p.13.

## Submitted journal articles

Gore, S.J., Franklyn-Miller, A., Richter, C., Falvey, E.C. King, E., and Moran, K.A., 2017. The effects of rehabilitation on the biomechanics of athletic groin pain patients. Journal of Medicine and Science in Sports and Exercise. Under review.

## Conference proceedings

### Oral presentations

Undheim, M., Gore, S.J., Franklyn-Miller, A., King, E., Boland, M., Carolan, D., Murphy, A., Falvey, E., 2014. Lower Extremity Isokinetic Muscle Strength and Joint Kinetics During Landing Tasks XXIII International Conference of Sport Rehabilitation and Traumatology. Milan,Italy.

Gore, S., Richter, C., Marshall, B.M., Franklyn-Miller, A.D., Falvey, E.C. and Moran, K.A., 2014. A Comparison Of Asymmetry In Athletic Groin Pain Patients And Elite Rugby Union Players Using Analysis Of Characterising Phases. International Society of Biomechanics in Sport. Tennessee,USA.

Marshall, B.M., Franklyn-Miller, A.D., Mealey,E., Gore, S., Moran, K.A., and Falvey, E.C., 2014. Can A Single Leg Squat Screen Provide An Insight Into Neuromuscular Control During A Single Leg Drop Landing And A Running Cut. International Society of Biomechanics in Sport. Tennessee,USA.

Gore, S.J., Marshall, B.M., Richter, C., Franklyn-Miller, A.D., Falvey, E.C. and Moran, K.A., 2014. Inter Limb Asymmetry in Athletic Groin Pain and Rugby Union Players. International Society of Biomechanics in Sport. Tennessee,USA.

Gore, S.J., Marshall, B.M., Richter, C., Franklyn-Miller, A.D., Falvey, E.C. and Moran, K.A., 2014. Asymmetry in athletic groin pain patients during a

lateral hurdle hop. Annual Conference Of World Federation Of Athletic Trainers. Dublin,Ireland.

Marshall, B.M., Franklyn-Miller, A.D., Moran, K.A., Richter, C., Strike, S., Gore, S., King, E., and Falvey, E.C., 2014. Biomechanical symmetry in elite injury free rugby union players during dynamic movements. Annual Conference Of World Federation Of Athletic Trainers. Dublin,Ireland.

Marshall, B.M., King, E., Gore, S., Holland, E., Franklyn-Miller, A., Moran, K.A., Strike, S., and Falvey, E.C., 2015. The Relationship between Hop Distance and Control in ACLR Patients. XXIV International Conference of Sport Rehabilitation and Traumatology Football Medicine Strategies for Player Care. London,United Kingdom.

Marshall, B.M., Richter, C., Gore, S., King, E., Halpin. R., West, S., Franklyn-Miller, A., Moran, K.A., Strike, S., and Falvey, E.C., 2015. Athletic Groin Pain: A Biomechanical Diagnosis. From Pain to Performance. London,United Kingdom.

Gore, S., Richter, C., Marshall, B.M., Franklyn-Miller, A.D., Falvey, E.C. and Moran, K.A., 2015. The Effects Of An Exercise Intervention On The Biomechanics Of An Athletic Groin Pain Cohort During The Lateral Hurdle Hop. International Society of Biomechanics in Sport. Poitiers, France.

Gore, S., Undheim, M., Marshall, B.M., Richter, C., King, E., Moran, K., Franklyn-Miller, A.D., Falvey, E.C. 2015. The biomechanical changes in cutting mechanics from 6 to 9 months post-primary ACL reconstruction. British orthopaedic association annual congress. Liverpool, United Kingdom.

Marshall, B.M., Moran, K., Franklyn-Miller, A., Falvey, E.C., King, E., Gore, S., and Strike, S. 2015. The use of 3d in motion capture in ACLR and athletic groin pain rehabilitation. From Pain to Performance. London,United Kingdom.

King, E., Richter, C., Gore, S., Strike, S., Franklyn-Miller, A., Falvey, E.C., Myer, G., Moran, R. 2016. Differences in biomechanical variables during single

and double leg landing between those that had a second ACL injury and those that did not after ACL reconstruction. British orthopaedic association annual congress. Belfast, Northern Ireland.

Richter, C., Gore, S., King, E., Strike, S., Franklyn-Miller, A., Falvey, E.C., Jackson, M., Moran, R. 2016. The use of machine learning techniques to predict re-injury risk from questionnaire data post ACL reconstruction. British orthopaedic association annual congress. Belfast, Northern Ireland.

Gore, S., Franklyn-Miller, A.D., Richter, C., Falvey, E.C., King, E., and Moran, K.A., 2017. Biomechanical complexity ? A measure to delineate between those with and without athletic groin pain. International Society of Biomechanics in Sport. Cologne,Germany.

Gore, S., Franklyn-Miller, A.D., Richter, C., Falvey, E.C., King, E., and Moran, K.A., 2017. Stiffness as an outcome measure of rehabilitation in athletic groin pain patients. IXXVI International Conference of Sport Rehabilitation and Traumatology. Barcelona,Spain.

**Poster presentations**

Gore, S., Richter, C., Marshall, B.M., Franklyn-Miller, A.D., Falvey, E.C. and Moran, K.A., 2014. The use of Analysis Of Characterising Phases to examine Athletic Groin Pain patients. Insight Student Conference. Dublin, Ireland.

Gore, S., Richter, C., Marshall, B.M., Franklyn-Miller, A.D., Falvey, E.C. and Moran, K.A., 2015. Change of direction biomechanics continue to improve from six to nine months post Anterior Cruciate Ligament reconstruction. Insight Student Conference. Galway, Ireland.

Welch, N., Falvey, E., Franklyn-Miller, A., Moran, K., Coyle, J., Antony, J., Gore, S., Richter, C., 2015. The effects of a free weight based strength-training intervention on pain, lumbar fat infiltration and biomechanics. Annual European Congress of Rheumatology. Rome, Italy.

Gore, S., 2016. An investigation into novel stiffness measures in athletic groin pain patients. Insight Student Conference. Dublin, Ireland.

Daniels, K., Franklyn-Miller, A., Richter, C., King, E., Gore, S., Moran, K., Faley, E., 2017. Changes in the biomechanics of a reactive cutting manoeuvre in an athletic groin pain cohort following a successful rehabilitation intervention. International Society of Biomechanics in Sport. Cologne, Germany.

# Contents

# List of Figures

# List of Tables

# Abbreviations and Acronyms

| | |
|---|---|
| **3D** | Three Dimensional |
| **ACP** | Analysis of characterising phases |
| **AGP** | Athletic groin pain |
| **CV** | Cross validated |
| **HAGOS** | The Copenhagen Hip and Groin Outcome Score |
| **HAGOS-FSR** | HAGOS Function in Sport and Recreation subscale |
| **HH** | Hurdle hop |
| **ICC** | Intraclass correlation coefficient |
| **QSE** | Quadratic sample entropy |
| **RBC** | Rank-biserial correlation |
| **RTP** | Return to play |
| **ROM** | Range of motion |
| **RMSE** | Root mean square error |
| **SEM** | Standard error of measurement |
| **SLDL** | Single leg drop landing |
| **SnPM** | Statistical non-parametric mapping |
| **SVM** | Support Vector Machine |

# Chapter 1

# Introduction

Groin Injuries are common in sports that involve rapid direction change. Indeed the incidence of groin injuries in male field sports is between 9.4 - 23 % of all injuries depending on the field sport played (Werner et al. 2009, Murphy et al. 2012, O'Connor 2004). Of these groin injuries 61 - 73 % are overuse in nature (Hölmich et al. 2013, Werner et al. 2009), and are typically characterised by an insidious onset. These overuse groin injuries herein termed Athletic Groin Pain (AGP), will be the primary focus of this PhD thesis. The high prevalence and morbidity associated with AGP has led to a number of literature reviews into the risk factors and treatment of AGP (Waldén, Hägglund and Ekstrand 2015, Serner et al. 2015, Whittaker et al. 2015, Esteve et al. 2015). Despite this, AGP remains poorly understood, potentially because of the anatomical complexity of the pubic region (Robertson et al. 2009). Critical to injury prevention and management is the ability to identify risk factors (via screening), implement effective injury prevention and rehabilitation programs and subsequently evaluate the success of those interventions. The anatomical complexity of the pubic region however, makes the evaluation of AGP difficult since neither pain (Orchard et al. 2000) nor radiological findings (Branci et al. 2015) are reliable methods of identifying the cause of the injury. An alternative to the traditional clinical or radiological

examination is the use of three-dimensional (3D) biomechanics in the assessment of an injury. Given that injuries are directly caused by relative excessive loading, biomechanical assessment incorporating an evaluation of technique and joint loading is an essential step for the understanding of injury risk factors and the subsequent development of injury prevention programmes (Bahr and Krosshaug 2005, Finch 2006).The need to better evaluate and understand the risk factors associated with AGP, is perhaps best substantiated by the high rate of reinjury and chronicity associated with this condition (Orchard et al., 2013 Falvey et al., 2016), suggesting that our understanding of AGP may be limiting our ability to effectively treat and screen for this injury

While 3D biomechanical assessments have been conducted quite extensively with other injuries [e.g. anterior cruciate ligament (ACL) (Lin et al. 2012) and lateral ankle ligament sprain (Willems et al. 2005)], there remains a dearth of literature in the area of AGP biomechanics. In fact prior to 2017 there was no research investigating the 3D biomechanics associated with this injury. Furthermore, outside of our research group, research in this area have utilised small sample sizes (n = 7-11) limiting the ability to draw robust conclusions regarding the biomechanics of AGP and warranting the further investigation of this injury. The primary aim of this PhD thesis is to evaluate the biomechanical factors affected by AGP to enhance our understanding of this injury.

## 1.1 Thesis Structure and Study Aims

This thesis incorporates a literature review (Chapter 2), the work from eight investigations including one published systematic review incorporated in the review of literature (Does the amount of lower extremity movement variability differ between injured and uninjured populations?) and two methodological studies investigating the number of trials required to obtain a stable representation

of a mean (Chapter 6) and variability (Appendix B). The five experimental studies included in this thesis are as follows:

**Chapter 3.** The effects of an exercise intervention on the biomechanics of an athletic groin pain group during a hurdle hop test.

Given that excessive relative loading causes injuries, an extensive 3D biomechanical analysis is a useful means of identifying potential risk factors for an injury. Despite this, there remains a dearth of literature examining the biomechanics of AGP and no research examining the biomechanics of the lateral hurdle hop in this population. This study examined 65 AGP patients and 50 uninjured controls during a lateral hurdle hop test using a continuous analysis approach. The aim of this study was to investigate the kinematic and kinetic variables that change in AGP patients after successful completion of an exercise intervention in comparison to a matched uninjured group. This study has been submitted to the journal of Medicine & Science in Sports & Exercise.

**Chapter 4.** Is stiffness related to athletic groin pain?

Stiffness has gained attention in recent literature as a potential modifiable risk factor for injury. Despite this, the association between stiffness and AGP has not been explored. Using the same dataset as Chapter 3, the aim of this study was to determine if AGP affects whole body vertical and joint stiffness and if so whether return to play following rehabilitation is associated with a change in stiffness. This study has been published in the Scandinavian journal of Medicine & Science in Sports.

**Chapter 5.** Biomechanical Complexity: A useful measure to delineate between those with and without athletic groin pain?

Complexity is a hallmark of physiological systems, with healthy organisms producing complex signals (Stergiou 2016). Despite this, there remains a dearth

of literature examining the association between injury and complexity, with no research investigating its association with AGP. This study examined complexity, as measured using a form of entropy, in 96 AGP male subjects compared to 50 uninjured male controls during a lateral hurdle hop test. The aim of this study was to examine the complexity of moment, angle and angular velocity waveforms in both AGP and uninjured populations during a lateral hurdle hop task.

**Chapter 7.** Cluster specific biomechanical changes post rehabilitation and their association with change in an outcome measure

Recently our research group have demonstrated for the first time the presence of biomechanical sub-clusters of AGP patients during a running cut task and have suggested that rehabilitation should be specifically tailored for each cluster (Franklyn-Miller et al. 2016). However, it is not known if these clusters would respond differently to a rehabilitation programme. The aim of this study was to quantify the relationship between the change in biomechanics and the change in 'pain and function' following rehabilitation within each cluster.

**Chapter 8.** Is movement variability during a running cut affected by athletic groin pain?

Movement variability in uninjured individuals represents the natural variation in movement patterns across multiple repetitions of the same task (Bernstein 1967). However there is divided opinion with respect to the relationship variability has with injury. Within the literature there is evidence of both greater and less variability associated with injury (Baida et al. 2017), while others have theorised that there may be an optimal level of variability. The primary aim of this study was to investigate if the magnitude of variability differed between those with and without AGP across the total waveform, and secondly to determine if within this

cohort there was any evidence of those without AGP exhibiting an optimal level of variability.

As a body of work, this thesis is the one of the largest investigations of AGP biomechanics conducted to date as it includes a 3D biomechanical assessment of over 200 AGP patients and 85 uninjured subjects. It is anticipated that this thesis will make substantial theoretical, empirical, methodological and practical contributions to both AGP research and clinical practice by enhancing our understanding of the biomechanical factors associated with this injury.

# Chapter 2

# Literature Review

## 2.1 The Anatomy of the Hip and Groin

To better understand the aetiology of AGP, an appreciation for the complex anatomy of the pelvis, hip and surrounding soft tissues is required. This section provides a brief overview of the anatomy involved with specific attention to the pubic symphysis and attaching structures. For a more thorough review of the anatomy of this region however, the reader is referred to other reviews (Robertson et al. 2009, Falvey, Franklyn-Miller and McCrory 2009, Hughes, Hsu and Matava 2002).

### 2.1.1 The Pelvic Girdle

The bony pelvis has two main functions of transferring weight from the upper body to the axial skeleton, and to withstand compression forces resulting from its support of body weight (Meyers, Greenleaf and Saad 2005). As a whole, four bones comprise the pelvis: two pelvic bones, the sacrum, and the coccyx. The two pelvic bones unite posteriorly with the sacrum at the sacroiliac joints and unite anteriorly at a joint called the pubic symphysis (Tortora and Derrickson 2008). Taking one of the pelvic bones from a lateral view, it consists of three

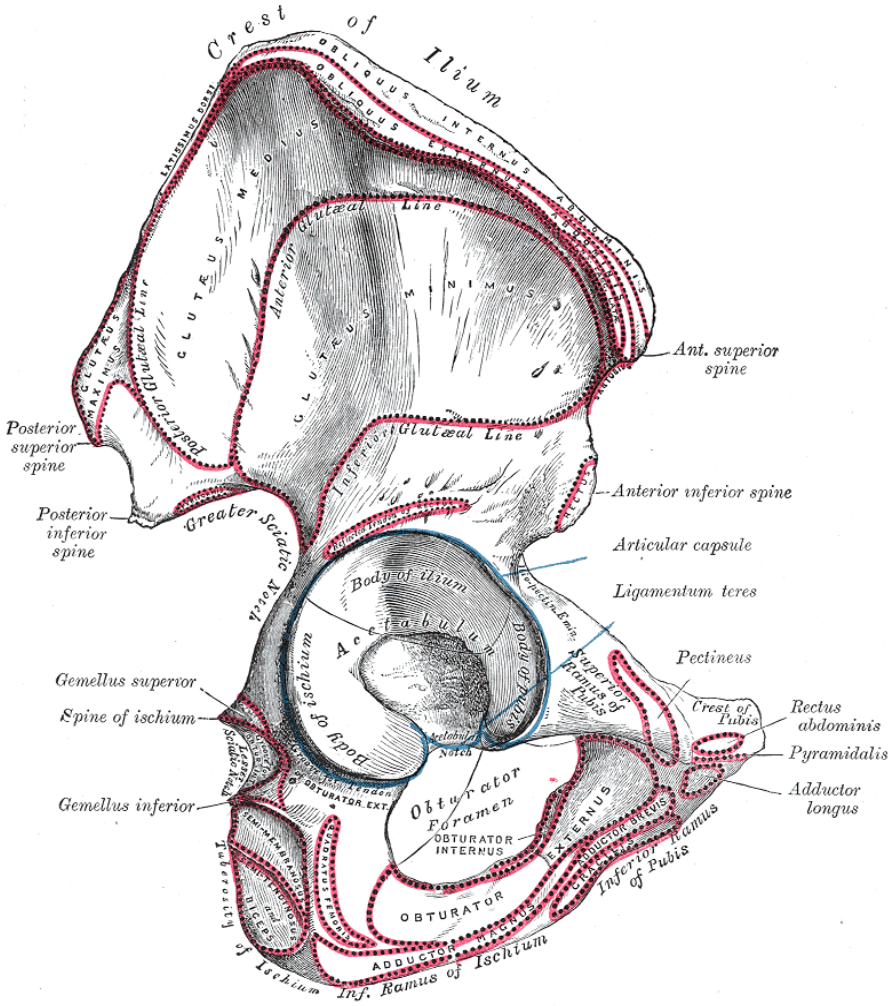


Figure 2.1: Surface of Pelvic bone (Gray and Lewis 1918)

sections namely the ilium, ischium and pubis. The ilium is the largest of the three sections and is most superior of the bones. The ischium lies in the inferior, posterior portion of the pelvic bone whilst the pubis is located in the inferior, anterior portion. Together, amongst other roles, the junction of the three sections helps form the acetabulum which acts as the socket that accepts the head of the femur to form the hip joint (Tortora and Derrickson 2008) (Figure 2.1).

### 2.1.2 Hip Joint

The hip joint (Figure 2.2) is ball and socket joint (spheroidal joint) consisting of the ball-shaped femoral head and cup-like depression of the acetabulum (pelvic

bone), which allows a wide range of motion in all directions. The femoral head is completely lined by articular cartilage, except for a small region on the central portion termed the fovea capitus (Macmahon et al. 2010). The ligamentum teres arises from this central depression before coursing inferiorly to insert onto the transversus ligament (Tortora and Derrickson 2008). The articular surface of the acetabulum is horseshoe-shaped, and lined by hyaline cartilage with a gap inferiorly (Blankenbaker and Tuite 2006). The transversus acetabular ligament traverses this non-articulating notch to support this part of the acetabular labrum (Hughes, Hsu and Matava 2002, Tortora and Derrickson 2008). Overlying the perimeter of the acetabulum is the fibro-cartilaginous labrum, the thickest portion of which lies posterosuperiorly (Macmahon et al. 2010). This functions to help stabilise the hip by deepening the acetabulum (Blankenbaker and Tuite 2006). The articular capsule inserts onto the acetabular rim directly, and the strong fibrous capsule encloses the hip joint to aid in the maintenance of hip stability (Hughes, Hsu and Matava 2002). The hip is further stabilised by capsular thickenings comprising the iliofemoral, pubofemoral and ischiofemoral ligaments (Macmahon et al. 2010).

### 2.1.3 Pubic symphysis and groin region

While contraction of antagonist musculature can result in increased agonist loading, most of the focus in AGP is on the anterior half of the pelvis given its proximity to the regions of pain and suspected injury in AGP. A key part of anterior pelvic anatomy that forms the fulcrum for many of the movements is the pubic symphysis (Meyers, Greenleaf and Saad 2005) (Figure 2.3). This structure is the joint between the two pubes of the pelvic bone and consists of a disc of fibrocartilage (Tortora and Derrickson 2008). The ligaments that stabilize this joint are the superior pubic ligament and inferior pubic ligament, however the muscles

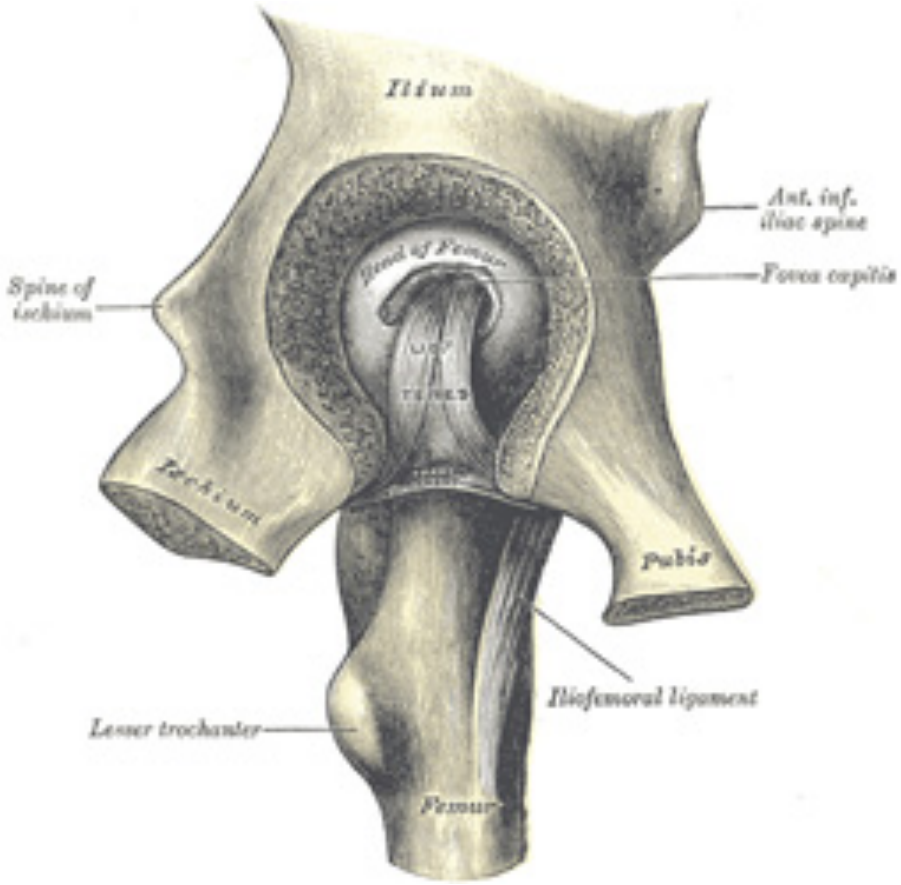


Figure 2.2: Hip Joint within the acetabulum (Gray and Lewis 1918)

attaching to the pubic symphysis probably play more of a role in stabilization of the joint (Meyers, Greenleaf and Saad 2005). Superiorly, the internal oblique, external oblique, and transversus abdominis muscles form the rectus sheaths, which enclose the rectus abdominis muscles, with fibres attaching to the superior pubic joint. The junction of where the abdominal structures converge at the pubic bone revolves around the inguinal canal, which contains the spermatic cord and ilioguinal nerve in males (Falvey, Franklyn-Miller and McCrory 2009). Inferiorly, the adductor longus tendon attaches to the anterior surface of the pubic body (Davis, Stringer & Woodley 2011). Anteriorly, fibres of the rectus sheath continue past its superior attachment and join with fibres from the adductor longus tendon extending up from its inferior attachment to form an aponeurosis covering the anterior surface of the pubic bone (Schilders 2000). The rectus abdominis and ad-

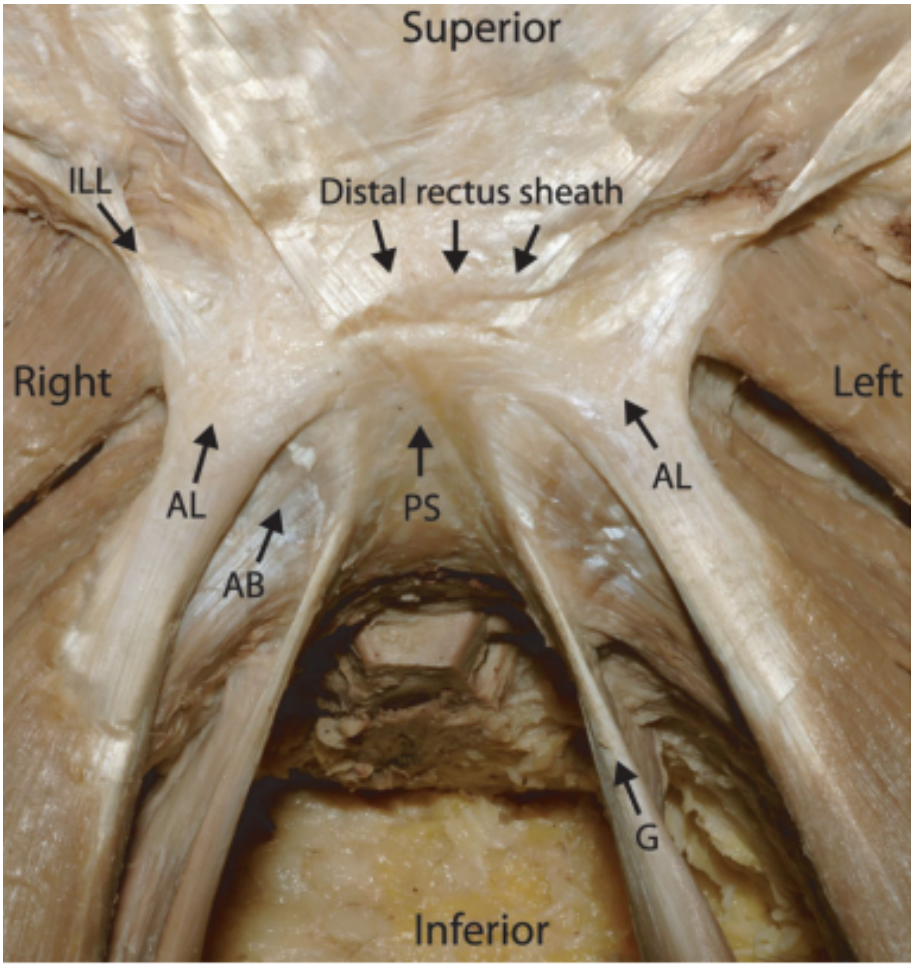


Figure 2.3: Cadaveric Dissection depicting the structures attaching into the pubic symphysis(PS). Adductor longus (AL); Ilio-inguinal ligament (ILL); Gracilis (G) and adductor brevis (AB)(Norton-Old et al. 2013)

ductor longus attach in continuity via the capsular tissues of the pubic symphysis (Robinson et al. 2007) and almost directly oppose each other (Meyers, Greenleaf and Saad 2005).

### 2.1.4 Pathogenesis

The above muscular structures contribute to a composite mechanism that provide pelvic stability and force transmission. This becomes important when considering the repeated loading of the pubic region commonly described in the aetiology of AGP. For example, it is thought that imbalances between abdominal and adductor muscle groups disrupt the equilibrium of forces around the symphysis

pubis resulting in overload of this region (Rodriguez et al. 2001). As noted by Robertson et al. (2009) the complex anatomical connections of the anterior pelvis provide a mechanism for the overlapping pathologies often described in the literature (Lovell 1995). Furthermore it explains why AGP often involves multiple structures, whereby the pain can be local, referred and may even radiate resulting in vague, diffuse symptoms and inconsistent clinical findings (O'Brien and Delaney 1997, Robertson et al. 2009). Indeed, a cadaveric study by Norton-Old et al., (2013) identified a clear anatomical continuity between the proximal adductor longus tendon, the pubic symphysis anterior capsule, and the distal rectus abdominis attachment. Using a strain gauge, the authors demonstrated that load applied to the adductor longus tendons resulted in strain measured in the ipsilateral and contralateral rectus sheaths. This provides a direct mechanism from which suprapubic pain can be experienced with increased tensile force from the adductors. Furthermore, cadaveric dissection has suggested that weakened rectus abdominis can affect pelvic alignment pressure directly over the adductor compartment (Meyers et al., 2000). Hip pathologies also often co-exist with AGP. Indeed one study identified that 92% of patients with tears of the acetabular labrum reported experiencing groin pain (Burnett et al. 2006). Furthermore Weir et al. (2011) identified that 94% of patients with groin pain demonstrated radiological signs of femoroacetabular impingement which may lead to hip pathologies and subsequently groin pain. These hip pathologies can ultimately result in referred pain and altered force transmission from the femur to the pelvis causing overload of the bony, muscular, and/or tendinous tissues of the anterior pelvis (Bowman, Fox and Sekiya 2010).

### 2.1.5 Terminology

Within the clinical literature related to AGP, there has been a wide range of terminology utilised to define and describe the clinical diagnosis underlying AGP. For example, a recent systematic review of 72 studies on the treatment of groin pain in athletes included 33 different diagnostic terms (Serner et al. 2015). This has added complications to clinical research investigating this already confusing area. Recently however, there was Doha consensus meeting to standardise the taxonomy of groin injuries in athletes (Weir et al. 2015). While not the focus of this current thesis, it is worth noting that the clinical diagnosis utilised within this current PhD does not fully follow this Doha agreement. As described in detail by Falvey et al. (2016), our research group identify an additional site of pain on palpation of the rectus aponeurosis on resisted sit up. This is not well differentiated in the Doha consensus statement and as such, this thesis does not fully comply with its proposed entities.

Furthermore, and of more relevance to this current thesis, Franklyn-Miller et al. (2016) contend that AGP is caused by an overload of the anterior pubic area (pubic symphysis and surrounding tissues), with various structures becoming painful in direct response to this loading or in an attempt to stabilize the region. Despite being diagnosed to different clinical entities, Franklyn-Miller et al., (2017), identified no association between the clinical diagnosis and movement pattern. Furthermore, following an exercise rehabilitation program common across all subjects, King et al., (2018) found no difference in time to return to play between subjects with different clinical diagnosis. In light of these findings, it is suggested that the identified painful structures during clinical examination may be of less importance than the painful propagative movements during dynamic activities. For this reason, while clinical diagnosis is reported within this thesis, AGP is utilised as an umbrella term for all clinical entities identified by Falvey et al.,

(2016) which are overuse in nature with a reported symptom duration greater than 4 weeks (Falvey et al., 2016). Within this thesis AGP is biomechanically examined as a whole group irrespective of sub clinical entities.

### 2.1.6 The clinical diagnosis of AGP

AGP is an umbrella term for the combination of one or more clinical presentations/diagnoses in the hip and groin region. These clinical diagnoses of AGP are typically conducted considering the athlete's history, clinical examination and radiological findings (Falvey et al., 2016).

Radiology can been used as a non-invasive insight into the pathology of underlying tissues and structures and has been identified as an important tool for the diagnosis of various injuries including AGP (Falvey et al., 2016). However, it has been suggested that relying on radiological imaging alone may result in a high false positive rate since asymptomatic soccer players often demonstrate what are considered to be abnormal MRI findings commonly associated with AGP (Branci et al., 2015). For example there are three radiological findings that are commonly reported in the pubic symphysis/ aponeurosis region with AGP (Chopra and Robinson, 2016). At the adductor origin, high signal intensity can help diagnose micro-tearing or separation of the adductor from the pubic bone. A secondary cleft sign is also common at the adductor longus insertion and is thought to represent a tenoperiosteal avulsion that occurs at the adductor longus tendon attachment. Finally, pubic bone oedema and degenerative changes of the pubic symphysis are also common presentations. Despite this, only higher grades of pubic bone marrow oedema, when present, could distinguish asymptomatic and symptomatic soccer players (Branci et al., 2015). For this reason, radiological imaging is recommended only in conjunction with clinical examination (Falvey, Franklyn-Miller and McCrory, 2009; Falvey et al., 2016). Falvey et al., (2016) note

that while MRI has low sensitivity for AGP, it has high specificity, particularly for the location of pain associated with adductor and pubic aponeurosis pathology. The following sections will outline some of the clinical and radiological findings identified in the examination of AGP examination and utilised within this thesis (Falvey, Franklyn-Miller and McCrory, 2009; Falvey et al., 2016).

**Pubic aponeurosis injury**

Pubic aponeurosis injury was the most commonly identified diagnosis in the study by Falvey et al., (2016). It is identified by pain on palpation of the rectus abdominis insertion into to the superomedial aspect of the pubic bone. Pain provocation tests include pain at the pubic insertion of the rectus abdominis during adductor squeezes and resisted abdominal contraction. These findings are supported by high MRI signal intensity at the pubic bone with micro-tearing or separation of the pubic aponeurosis from the pubic bone.

**Adductor injury**

Adductor injury is defined by tenderness at the adductor origin on the inferior aspect of the pubic bone. Pain is experience at the adductor origin on squeeze test and passive abduction of the hip. MRI findings may include a high signal intensity at the pubic bone with micro tearing or separation of the adductor from the pubic bone.

**Hip Injuries**

Hip Injuries are diagnosed via reduced ROM at the hip, pain on FABER and/or FADIR tests and MRI findings consistent with hip joint pathology (e.g. cam and/or pincer morphology). Both Falvey et al., (2016) and Thorborg et al., (2018)

caution against utilising imaging as the main guide for treatment in hip pain due to the high false positive rate possible.

#### Hip flexor injury

Hip flexor injury is diagnosed via pain at the iliopsoas muscle that is worsened with resisted hip flexion and passive stretch. While MRI is not used within this thesis to help diagnose hip flexor injury, Thorborg et al., (2018) suggest it may be helpful to improve the accuracy of the initial diagnosis. MRI findings may include fluid distension anterior to the hip joint capsule indicating iliopsoas bursitis (Chopra and Robinson, 2016).

#### Inguinal injury

Finally, Inguinal injury is diagnosed via pain on palpation of the region, where clinical and imagining findings outlined above are not identified and resolution of symptoms occurs under local anaesthetic infiltration of the ilioinguinal canal.

### 2.1.7 Summary

The above sections provided a brief overview of the anatomical structures involved in AGP. The complex and interconnected nature of the region is likely the reason AGP remains poorly understood. In particular the direct anatomical connections between adductor longus, rectus abdominis, obliques and transversus abdominis mean that it is unlikely that pain provocation or stress tests load single anatomical structures in isolation. As such the validity of a sports medicine practitioner to precisely diagnose isolated structures associated with chronic groin pain is questionable (Robertson et al., 2009). AGP is considered to be the combination of one or more clinical presentations/diagnoses in the hip and groin region. Within this thesis, clinical diagnosis is conducted considering the athlete's

history, clinical examination and radiological findings (Falvey et al., 2016) and includes the presentations of pubic aponeurosis injury, adductor injury, hip injury, hip flexor injury and inguinal injury. An alternative to this traditional clinical examination and radiological assessment is the use of 3D biomechanics for the assessment of AGP.

## 2.2 The Epidemiology of Athletic Groin Pain

Information regarding AGP epidemiology is the first essential step of an injury prevention program (Bahr and Krosshaug 2005, Finch 2006, van Mechelen, Hlobil and Kemper 1992). Whilst there are numerous epidemiology studies conducted on groin injuries, their inconstancy of terminology and inclusion criteria make cross comparison difficult (Hölmich and Thorborg 2014). For example some studies report hip and groin injuries combined (Werner et al. 2009) whilst others distinguish between these structures (Hölmich et al. 2013). Furthermore some research distinguishes between chronic and acute groin pain, whilst others do not (Hölmich and Thorborg 2014). In general, a universally accepted definition of an injury was only agreed upon in soccer in 2006 (Fuller et al. 2006). Despite this, most injury studies still require a time loss injury in order to be registered (Ekstrand, Hägglund and Waldén, 2011). This is problematic as its application depends on the frequency of training and games (Dvorak and Junge, 2000). Specifically with AGP an injured athlete might still participate in games but his performance may be hindered or the athlete may manage his symptoms around matches (Hölmich and Thorborg, 2014). In this respect the incidence of groin pain is likely under reported. With these limitations in mind, the following section will detail the incidence, severity and mechanism of injury in AGP currently reported in the literature.

### 2.2.1 Incedince of Injury

The incidence of injury, also known as the injury rate, is the number of new injuries in a specific period divided by the total number of players exposed to injury (van Mechelen, Hlobil and Kemper, 1992) and is critical feature of epidemiology research (Chambers, 1979; Wallace and Clark, 1988). AGP has been commonly reported in field sports involving repetitive twisting, turning and kicking (Quinn, 2010). In male soccer for example, according to a recent UEFA injury survey, hip/groin injury accounts for 12 - 16% of all injuries, with an absolute incidence rate of 1.1/1000 hours of play [3.5/1000 match hours and 0.6/1000 training hours] (Werner et al., 2009). In male inter-county Gaelic Athletic Association (GAA) football and hurling, groin injuries account for 9.4% and 10.3 % of all injuries respectively (Murphy et al., 2012; Blake et al., 2014). In rugby league groin injuries account for 23% of all injuries with an absolute incidence rate of 2.4/1000 hours (O'Connor, 2004). Groin injuries are less common in female athletes and account for only 4% of all injuries in female club level GAA football (Brown, Papadopoulos and Pritchett, 2013), and female elite football (Hägglund, Waldén and Ekstrand, 2009). Indeed results from a recent systematic review demonstrate that males are at a 2.0 times greater relative risk of developing a groin injury in soccer in comparison to females (Orchard, 2015).

Absolute groin injury incidence rate is also lower in sub-elite football populations (0.40 -0.80 injuries per 1000 hours) in comparison to elite players (Hölmich et al., 2013). The greater risk associated with higher level of play may result from a higher intensity in training and game play as well as a greater number of training hours (Hölmich et al., 2010). Alternatively the reported higher incidence may be simply due to a more efficient injury-reporting regimen available at elite clubs. In addition to the frequent occurrence of AGP, those with a history of groin pain are at a 2.4 times greater risk of redeveloping the injury (Hagglund, 2006). Whilst

groin injury incidence is clearly high in male sports, it is likely that the actual prominence of this injury is underreported. In order for an injury to be recorded in an injury registry it typically requires time loss from matches (Werner et al., 2009). The chronic nature of AGP means that players often manage their pain around match play (Tyler et al., 2010). Indeed in two studies, 67-69% of players in soccer had reported some pain in this region during the soccer season (Hanna et al., 2010; Kristian Thorborg et al., 2011).

### 2.2.2 Severity of Injury

Injury severity is defined according to the length of incapacity (Sandelin et al., 1988). It is usually calculated as time lost from training and games through injury (van Mechelen, Hlobil and Kemper, 1992). Research from the UEFA league identified that mean absence from groin pain was $14 \pm 19$ days on the first incident then subsequently for reinjures as $23 \pm 27$ days (Werner et al., 2009). In contrast to this, research from sub elite soccer (Hölmich et al., 2013) demonstrated a median absence of 16 days with a range from 1 - 208 days. Of these injuries, 33% were characterised as severe (injury time $>$28 days). This large range of time loss may demonstrate the potential limitation of including both acute and overuse injuries in the single investigation since overuse groin pain or AGP is typically considered to have a high morbidity associated with it.

### 2.2.3 Mechanism of Injury

Mechanisms of injury are commonly divided into either contact or non-contact (van Mechelen, Hlobil and Kemper, 1992; Bahr and Krosshaug, 2005). This information is important as it highlights how preventable injuries may be. The exact mechanism of developing groin pain remains poorly understood (Falvey, King and Kinsella, 2015). This is problematic since it has been suggested that

a precise description of the inciting event is a critical to preventing injuries in sport (Bahr and Krosshaug, 2005). Whereas acute injuries are characterised by a specific injury event which can be characterized and examined, AGP is typically characterised by a insidious onset (Falvey, Franklyn-Miller and McCrory, 2009). For this reason, researchers have turned to the epidemiological perspective, to get a better understanding of how AGP occurs (van Mechelen, Hlobil and Kemper, 1992).

In light of this, while an exact specific injury event is unknown, it is generally accepted that AGP is a non contact injury caused by a chronic overload of the muscle, tendinous and bony structures of the anterior pelvis (Verrall et al., 2007). Indeed between 61 - 73% of groin injuries in soccer are overuse in nature (Werner et al., 2009; Hölmich et al., 2013) and most commonly associated with sports involving kicking, twisting, rapid acceleration and deceleration (Murphy et al., 2012; Thorborg, Rathleff and Petersen, 2015) . It has previously been suggested that kicking may be an important factor in the development of AGP, potentially between 30% and 45% of the swing phase while the adductor longus is both eccentrically most active and stretching most rapidly (Charnock et al., 2009). However it is likely that this mechanism is more closely linked to acute adductor strains than AGP. Furthermore, sports that involve little kicking (e.g. Gaelic Hurling) still demonstrate a high prevalence of AGP and the association of leg dominance and site of pain remains unclear (Werner et al., 2009; Hölmich et al., 2013). This may suggest that twisting and turning actions may be more involved in the mechanism of this injury (Blake et al., 2014) than kicking actions.

### 2.2.4 Summary

Epidemiological research has documented the high prevalence of the AGP in field-based sports. It is the third most common injury in soccer and Australian

rules football (Werner et al., 2009; Orchard, Seward and Orchard, 2013) and has high tendency to reoccur (Hagglund, 2006). Considering the high incidence of AGP, the evidence discussed substantiates the requirement for more effective injury prevention strategies. As suggested by van Mechelen, Hlobil, & Kemper, (1992), following establishment of the epidemiology surrounding an injury, the next step requires a detailed investigation into the risk factors associated with it. The following chapter will attempt to present the literature surrounding the risk factors for AGP.

## 2.3 The Risk Factors for Athletic Groin Pain

As noted by recent reviews (Maffey and Emery, 2007; Ryan, DeBurca and Creesh, 2014; Whittaker et al., 2015) a number of modifiable and un-modifiable risk factors have been proposed for groin injury. These reviews do not solely examine AGP however, but rather look at groin injury per se. As referred to in Section 2.1.5 of this thesis, AGP specifically refers to chronic overuse groin injuries typically characterised by an insidious onset. As such this section will primarily present research investigating prolonged groin pain with an injury definition including a time loss of greater than 2 weeks. Given the focus of this PhD on AGP biomechanics and the paucity of research in this area, an exception is made for studies investigating the biomechanics of AGP even without a well-defined time loss requirement. Results from this review are summarised in Table 2.1 on page 32. This section will review in detail the findings from 3D biomechanical studies of AGP before reviewing in brief identified risk factors categorized as 'measures of hip range of motion', 'hip strength/ activation', 'abdominal muscle function', 'body mass' and 'other measures'.

### 2.3.1 The 3D biomechanical risk factors for Athletic Groin Pain

At the time of starting this PhD (October 2013) there was no research investigating the 3D biomechanics of AGP. In recent years however there has been a growing interest in this area and subsequently in the last year, four studies have been published investigating the influence of AGP on the biomechanics of various athletic tasks (Edwards, Brooke and Cook, 2017; Janse van Rensburg et al., 2017; Severin et al., 2017; King et al., 2018). The following section will review the four studies in general before exploring their findings at a section/joint level. While these studies provide valuable information regarding what biomechanical factors may be affected by AGP, it is important to note that due to their retrospective nature, causation cannot be determined from these studies.

Given the close proximity of the hip and pelvis to the region of pain in AGP, it is not surprising that all biomechanical studies reviewed examined these two segments/joints. In fact both Janse van Rensburg et al., (2017) and Severin et al., (2017) solely examined the hip and pelvis in their investigation of landing and kicking, respectively. Within the study by Janse van Rensburg et al., (2017) the two greatest effect size differences were for increased hip abduction (Cohen's D: 1.12) and decreased pelvic contralateral tilt (Cohen's D: 0.75) at initial contact in the AGP group in comparison to the uninjured controls during a single leg landing. This finding is surprising since the femur relative to the pelvis defines the hip joint angle and a reduction in pelvic contralateral tilt typically results in a concomitant increase in hip adduction. Regardless, caution is required when interpreting the findings of this study since the wide range of participant ages (19 - 54) may have confounded the results observed.

Severin et al., (2017) investigated a 45° and 60° kicking task in 11 AGP subjects in comparison to 11 matched controls. During the 45° kick, the authors identified reduced pelvic range of motion and velocity for the sagittal and transverse plane in

the AGP group in comparison to the uninjured group. At the hip, reduced range of motion and flexion velocity was observed in the swing leg, while in the stance limb increased transverse range of motion and decreased flexion velocity were seen in the AGP group in comparison to uninjured controls. When examining the 60° kicking task, the authors identified similar trends (Table 2.1). The results from this study perhaps suggest that the AGP group are utilising a more restricted technique to avoid loading on anterior aspects of the hip and pelvis. This may explain the reduced pelvis and hip velocities observed since lower ROM would result in less pre-stretch of the anterior structures of the hip (Lees et al., 2010) which is vital to the enhancement of concentric power and force (Komi and Bosco, 1978; Harrison, Keane and Coglan, 2004). Indeed in support of this, the two greatest effect size observed in Severin et al., (2017) were for decreased pelvic posterior tilt velocity (Cohen's D: 0.87) and sagittal plane hip ROM (Cohen's D: 0.74). Interestingly, unlike the uninjured group, the AGP group demonstrated limited adaptations to the different approach angles suggesting reluctance, or decreased capacity, to alter the kinematics to the task constraints. As per the dynamic systems theory, this lack of adaptability could represent either a predisposition to AGP due to repetitive loading or alternatively that the pain is providing an organismic constraint to avoid loading the painful regions of the anterior pelvis (Hamill et al., 1999).

This concept of movement variability in AGP was specifically examined by Edwards, Brooke and Cook, (2017). The authors investigated the effects of AGP on cutting mechanics during an unanticipated 45° running cutting task in seven subjects with AGP in comparison to 10 matched controls. For variability, the authors examined 59 variables and identified in four variables (T12-T1 rotation at initial contact and peak vertical GRF, hip and knee rotation at weight acceptance) increased variability in the AGP group in comparison to uninjured controls. Interestingly, rather than using null hypothesis testing as their primary statistical technique, the authors used magnitude based inference (Batterham and Hopkins,

2006). Using this less conservative approach (in comparison to null hypothesis testing with an alpha level of 0.05) the authors identified a clear trend for decreased variability in the AGP group in comparison to uninjured controls for 53% (31/59) of the variables examined. Given these contrasting findings and the small sample size used by Edwards, Brooke and Cook, (2017), further research is require to substantiate their findings with respect to variability and AGP. In addition to examining variability, Edwards, Brooke and Cook, (2017) also investigated the kinematics of the cutting task. The authors found increased T12-T1 ipsilateral rotation at both initial contact and peak vertical GRF and also decreased hip internal rotation and knee external rotation during weight acceptance in those with AGP in comparison to uninjured controls. While it is unclear why, it is worth noting that the four significant different kinematic variables identified in this study were the same four variables identified in the examination of variability. Within this study, effect sizes were not explicitly defined for the kinematic findings. However for the variability results, the greatest effect sizes were demonstrated by T12-T1 rotation at initial contact (Cohen's D: 1.17) and peak vertical GRF (Cohen's D: 0.99).

The final study investigating the biomechanical risk factors of AGP is from our research group (King et al., 2018). This study differs to the previous three studies in two main ways. Firstly in comparison to the other research, King et al., (2018) utilises a pre-post cohort study design. This is a different and much less common approach to identifying potential risk factors for injury than case control examinations (as used in the other three studies reviewed here). As per the probabilistic approach to causation (Burr, 2003; Marshall and Moran, 2015), the biomechanical factors that change with a successful rehabilitation are possibly causative of the improvements in injury status and therefore likely associated with the injury. Using this approach King et al., (2018) examined the biomechanics of 112 AGP patients pre- and post- a successful intervention and

return to sport. The authors identified multiple kinematic and kinetic changes from pre- to post- rehabilitation with effect sizes ranging from 0.26 to 0.79. The greatest overall effect sizes were for the HAGOS questionnaire (Cohen's D: 0.59 - 1.78), which was an outcome measure of the rehabilitation program. Within the biomechanical changes however, the highest effects sizes were seen for reductions in ipsilateral trunk side flexion (Cohen's D: 0.79) and increased contralateral pelvic rotation in the direction of travel (Cohen's D: 0.76) from pre- to post- rehabilitation. Interestingly, the biomechanical changes observed in this study also corresponded with reduced work at the hip in all three planes, perhaps suggesting reduced loading in the surrounding region. The following section will explore the findings of these four studies at a joint/segment level.

**Thorax mechanics**

Two studies have investigated the influence of AGP on thorax mechanics during a cutting action (Edwards, Brooke and Cook, 2017; King et al., 2018). King et al., (2018) identified increased contralateral flexion (Cohen's D: 0.62 - 0.79) and decreased ipsilateral rotation (Cohen's D: 0.46 - 0.54) from pre-post rehabilitation both absolutely and relative to the pelvis for the majority of the ground contact. Similarly Edwards, Brooke and Cook, (2017) also found increased T12-T1 ipsilateral rotation at both initial contact and peak vertical GRF in those with AGP in comparison to uninjured controls (Cohen's D: $\leq 0.80$).

**Pelvis mechanics**

All four studies investigating the influence of AGP on movement biomechanics have examined the pelvis. Two studies investigated range of motion at the pelvis (Janse van Rensburg et al., 2017; Severin et al., 2017) however of these, only Severin et al., (2017) identified a significant difference between those with and without AGP. The authors identified increased range of motion and decreased angular

velocity in the AGP group in comparison to uninjured controls for both the sagittal and transverse plane of the pelvis during a kicking task (Cohen's D: 0.52 - 1.12). Neither study identified frontal plane range of motion as a potential risk factor for AGP. When examining pelvis angles, three studies have examined the affects of AGP in all three planes (Edwards, Brooke and Cook, 2017; Janse van Rensburg et al., 2017; King et al., 2018), although Edwards, Brooke and Cook, (2017) examined the L5-S1 segment angles rather than the pelvis. Of these three studies, no research identified significantly different sagittal plane pelvis mechanics. In the frontal plane, two studies found that pelvis mechanics were affected by AGP (Janse van Rensburg et al., 2017; King et al., 2018), but with contrasting findings. During a single leg drop landing, Janse van Rensburg et al., (2017) found decreased pelvic contralateral tilt at both initial contact and the lowest vertical position in those with AGP in comparison to uninjured controls (Cohen's D: 0.35-0.75). In contrast King et al., (2018) found reduced contralateral tilt from pre- to post- rehabilitation (pre- rehab contralateral tilt $>$ post-rehab contralateral tilt) for the total ground contact (Cohen's D: 0.62). In the transverse plane, both Janse van Rensburg et al., (2017) and King et al., (2018) found increased pelvis ipsilateral rotation in those with AGP in comparison to uninjured controls (Cohen's D: 0.62) and post rehabilitation (Cohen's D: 0.76).

**Hip Mechanics**

Two studies have examined the range of motion at the hip in all three planes (Janse van Rensburg et al., 2017; Severin et al., 2017). In the transverse plane both studies identified increased range of motion in those with AGP in comparison to uninjured controls (Cohen's D: 0.43 - 0.52). In contrast, only Severin et al., (2017) found decreased sagittal plane ROM in the AGP group in comparison to uninjured control in the swing leg during a kicking task (Cohen's D: 0.60). Neither study found frontal plane range of motion as a potential risk factor for AGP. Three

studies have examined the affects of AGP on hip angles (Edwards, Brooke and Cook, 2017; Janse van Rensburg et al., 2017; King et al., 2018). In the sagittal plane only King et al., (2018) identified increased hip flexion pre- rehabilitation in comparison to post-rehabilitation [Hip flexion decreased significantly pre- to post- rehabilitation] for 0 -100% of ground contact (Cohen's D: 0.58). In the frontal plane both Janse van Rensburg et al., (2017) and King et al., (2018) found increased hip abduction in those with AGP in comparison to controls (Cohen's D: 1.12) and post- rehabilitation mechanics (Cohen's D: 0.36), respectively. In the transverse plane both Edwards, Brooke and Cook, (2017) and Janse van Rensburg et al., (2017) found decreased hip internal rotation at weight acceptance (Cohen's D: $\leq$ 0.80) and initial contact (Cohen's D: 0.60), respectively. Two studies investigated hip kinetics in all three planes (Edwards, Brooke and Cook, 2017; King et al., 2018), of these only one study found significant differences. King et al., (2018) found a decrease in hip adductor moments (Cohen's D: 0.39) from 78 to 95% of the ground contact, and a decrease in extensor moments (50-89% of the ground contact) from pre- to post- rehabilitation (Cohen's D: 0.39). The authors also identified an decrease in sagittal plane concentric power (Cohen's D: 0.43) from 68 to 87% of the ground contact and decreased work in all three planes (Cohen's D: 0.24 - 41) pre- rehabilitation in comparison to post- rehabilitation.

**Knee mechanics**

Two studies examined the of effects of AGP on both knee angles and kinetics in all three planes (Edwards, Brooke and Cook, 2017; King et al., 2018). In the sagittal plane, King et al., (2018) found a decrease in flexion from 57-100% of ground contact (Cohen's D: 0.33) from pre- to post- rehabilitation. Neither study indented any differences in the frontal plane. In the transverse plane Edwards, Brooke and Cook, (2017) found decreased external rotation at weight acceptance (Cohen's D: $\leq$ 0.80). Only King et al., (2018) identified difference in knee kinetics. In the

sagittal plane there was an increase in knee power generation (Cohen's D: 0.4) from pre- to post- rehabilitation (43 to 58% of the ground contact). There was also a decrease in total work done at the knee (Cohen's D: 0.26) and specifically in the frontal plane (Cohen's D: 0.36). No differences were identified in the transverse plane.

**Ankle mechanics**

Two studies examined the of effects of AGP on both ankle angles and kinetics in all three planes (Edwards, Brooke and Cook, 2017; King et al., 2018), however only King et al., (2018) identified between group differences. From pre- to post-rehabilitation, in the sagittal plane, King et al., (2018) identified an increase in dorsi-flexion from 9 to 75% of ground contact (Cohen's D: 0.58). No differences were identified in the frontal or transverse plane. In the sagittal plane there was an increase in planter flexor moment from 6 to 71% of the ground contact (Cohen's D: 0.48), an increase in eccentric power from 1 to 24% (Cohen's D: 0.46), an increase in concentric power from 57 to 83% (Cohen's D: 0.46) and an increase in sagittal plane work (Cohen's D: 0.70). In addition, there was an increase in total work done at the ankle from pre- to post- rehabilitation (Cohen's D: 0.68). No differences were identified in the frontal or transverse plane.

**Summary of the 3D biomechanical risk factors for Athletic Groin Pain**

The four studies investigating the 3D biomechanical factors associated with Athletic Groin Pain, suggest that the mechanics of the trunk, hip and pelvis may be of importance to AGP, however the identification of any clear trends is challenging given the small number of studies and the heterogeneity in study methodologies. In the absence of a large number studies investigating the 3D biomechanics of AGP, the findings from clinical examinations may be an important source of information to understand the risk factors for AGP. The following sections will explore

some of the clinical and anthropometric measurements explored in the literature with respect to AGP.

### 2.3.2 Measures of Hip range of motion

Restricted range of motion assessed during clinical examination has long been associated with AGP. Indeed the earliest reference of this association was in 1964 (Howse, 1964). Since then there has been a multitude of research examining its association to AGP. Two of the four studies reviewed identified reduced hip range of motion as a risk factor for AGP (Verrall et al., 2007; Nevin and Delahunt, 2014) while two did not (Malliaras et al., 2009; Edwards, Brooke and Cook, 2017). As suggested by Malliaras et al., (2009), this conflicting result may be explained (at least for their study) by inadequate statistical power. Indeed both Malliaras et al., (2009) and Edwards, Brooke and Cook, (2017) examined small sample sizes (AGP n = 10 and 7 respectively). Whilst Nevin & Delahunt (2014) identified multiple restrictions at the hip in AGP patients (Int/Ext Rotation, bent knee fall out ROM), Verrall et al., (2007) only identified prospectively that decreased total hip rotation ROM was associated with AGP. This discrepancy may be due to study design (cross sectional vs. prospective) and/or the sample size in the studies.

### 2.3.3 Hip strength/ activation

Since pain in AGP is typically experienced surrounding the hip and pelvis, muscle strength and activation may have a vital role to play in terms of tissue integrity and control of motion and forces in this complex anatomical location (Holmich et al., 1999). Decreased levels of hip adductor strength has been suggested as a risk factor for AGP since it is a commonly painful structure and decreased neuromuscular capacity would predispose this group of muscles to injury (Nevin & Delahunt, 2014). Four studies investigated hip adductor strength as a risk

factor for developing AGP (Malliaras et al., 2009; Crow et al., 2010; Nevin and Delahunt, 2014; Edwards, Brooke and Cook, 2017). Three out of four of these studies identified reduced adductor strength in those with AGP in comparison to controls (Malliaras et al., 2009; Crow et al., 2010; Nevin and Delahunt, 2014). In particular, one study identified prospectively a relationship between bilateral adductor strength reduction and the onset of AGP with decreased strength being identified up to two weeks prior to onset of pain (Crow et al., 2010). Hip abductor strength and activation has also been suggested to be important in the development of AGP since decreased function of these muscle may result in inefficient load transfer, altered stress across the pubic symphysis and the development of pain in associated structures (Morrissey et al., 2012). Three studies investigated hip abductor strength or activation in this review (Malliaras et al., 2009; Morrissey et al., 2012; Edwards, Brooke and Cook, 2017). Of these three studies, one study identified hip abductor strength or activation to be associated with AGP (Morrissey et al., 2012).The authors identified a lower gluteus medius to adductor longus activation ratio in AGP patients suggesting that these muscles should be addressed in the rehabilitation and prevention of AGP. Interestingly this was due to a reduction in hip abductor rather than adductor activation, potentially contradicting the findings of adductor weakness as a risk factor for AGP (Malliaras et al., 2009; Crow et al., 2010; Nevin and Delahunt, 2014). These opposing findings need to be explored further to ascertain the relationship between muscle strength and activation in patients with AGP.

### 2.3.4 Abdominal Muscle Function

Since the internal and external obliques, rectus abdominus and transversus abdominus join the fascia of the common adductors inferiorly to form an aponeurosis of the pubic symphysis (Robertson et al., 2009), any alteration of how these

muscles function may cause overload in these commonly painful structures in AGP. In particular, the transversus abdominus muscle is considered to play an important role in contributing to active stability of the pelvis and spine prior to movement of the lower body limbs (Aruin and Latash, 1995; Hodges and Richardson, 1997).

When exploring the timing of muscular activation using electromyography (EMG), Cowan et al., (2004) identified a delayed onset of the transversus abdominus during an active straight leg raise task. An alternative to using EMG as a measure of muscular activity is the use of muscle thickness as measured by ultrasound. Using this method Jansen et al., (2010) demonstrated a reduction in transversus abdominus resting thickness but not during an active straight leg raise. The contrasting findings can be explained by recent investigations indicating that muscle thickness may not be a valid surrogate for EMG activity in the abdominal muscles (Brown and McGill, 2010; Tahan et al., 2013). Regardless, the clinical relevance of transversus abdominus activity as a causative factor for AGP is questionable, since later research identified that artificially induced acute groin pain also reduces relative transversus abdominus thickness (Jansen et al. 2010). Cowan et al., (2004) also identified no association between the EMG onset of the oblique muscles or the rectus abdominus in patients with AGP.

### 2.3.5 Body Mass

Three studies investigated body mass as a risk factor for AGP (Verrall et al., 2007; Edwards, Brooke and Cook, 2017; Janse van Rensburg et al., 2017). Of these studies, one investigation prospectively identified lower weight as a risk factor for AGP in Australian rules football (Verrall et al., 2007). It is uncertain why this association was found, but may be due to higher agility demands of certain player positions. AGP is typically associated with agility type movements (running,

cutting, twisting) (Hölmich and Thorborg, 2014; Ryan, DeBurca and Creesh, 2014; Franklyn-Miller et al., 2016; Edwards, Brooke and Cook, 2017; King et al., 2018) and previous research has associated body mass with performance in agility tasks (Chaouachi and Brughelli, 2009). It is possible therefore that lighter athletes place higher relative loading on their bodies through completion of athletic agility movements at a higher velocity. Conversely, whilst conducted prospectively (Verrall et al., 2007), the small number of players who subsequently developed AGP (n = 4) also means the external validity of the study was low.

### 2.3.6 Other Measurements

Of the other risk factors investigated in this review, HAGOS scores and pain during clinical tests were significantly associated with AGP (Cowan et al., 2004; Jansen et al., 2010; Nevin and Delahunt, 2014; King et al., 2018). These associations whilst important for diagnostic and tracking purposes are a clearly a cause of AGP rather being a preceding risk factor. They also highlight the weakness of assessing AGP patients retrospectively for the identification risk factors.

Table 2.1: Summary of risk factor research

| Author | Study Design | Participants | Examined Variables | Significant Findings |
|---|---|---|---|---|
| **Verrall et al., 2007** | Prospective Cohort | Elite Male Australian football<br><br>N = 29 (4 developed groin pain)<br><br>Age: Mean (range): 21 (18-30)<br><br>Pain for > 6 wks. | • Age<br>• Height<br>• Weight<br>• Hip Internal rotation<br>• Hip External rotation<br>• Total hip rotation | • Lower weight<br>• Less total hip rotation |

Table 2.1: Summary of risk factor research (cont.)

| | | | | |
|---|---|---|---|---|
| **Nevin & Delahunt 2014** | Case-control (matched design) | Club-level Male Gaelic football<br><br>N = 36 (18 with groin pain)<br><br>Age: Mean (SD): Injured 23.9 yrs. (3.2) Uninjured 23.8 (3.6)<br><br>Pain for > 6 wks. | • Hip Internal rotation<br>• Hip External rotation<br>• Bent knee fall out ROM<br>• Hip adduction squeeze test (mmHg) 45° of hip flexion<br>• HAGOS Questionnaire | • Hip Internal rotation<br>• Hip External rotation<br>• Bent knee fall out ROM<br>• Hip adductor squeeze<br>• HAGOS |

Table 2.1: Summary of risk factor research (cont.)

| | | | | |
|---|---|---|---|---|
| **Morrissey et al., 2012** | Case-control (matched design) | Male Football code<br><br>N =18 (9 with groin pain)<br><br>Age: Mean (range): 25 yrs. (18- 35)<br><br>Pain for > 4 wks. | **During single leg standing hip flexion:**<br><br>• Gluteus medius to adductor longus activation ratio | • Lower gluteus medius to adductor longus activation ratio |

Table 2.1: Summary of risk factor research (cont.)

| | | | | |
|---|---|---|---|---|
| **Jansen et al., 2010** | Case-control | Male Soccer, running, field hockey<br><br>N = 65 (42 with groin pain)<br><br>Age: Mean (SD): 25.6 (7.3)<br><br>Pain for > 6 wks. | • Pain (Likert 10) during an active straight leg raise and hip adduction<br>• Thickness of transversus abdominis and internal obliques at rest and during ASLR and hip adduction. | • Pain during active straight leg raise.<br>• Transversus abdominis resting thickness. |

Table 2.1: Summary of risk factor research (cont.)

| | | | | |
|---|---|---|---|---|
| **Crow et al., 2010** | Prospective cohort | Elite junior soccer (sex not reported)<br><br>N = 86 (12 developed groin pain)<br><br>Age: 16- 18 yrs.<br><br>Pain for > 2 consecutive wks. | • Adductor strength | • Reduced adductor strength |

Table 2.1: Summary of risk factor research (cont.)

| | | | | | |
|---|---|---|---|---|---|
| **Cowan et al., 2004** | Case-control | Male Australian football<br><br>N=22 (10 with groin pain)<br><br>Age = Mean (SD): 26yrs (7)<br><br>Pain for > 6 wks. | • Pain (Visual Analog Scale)<br><br>**During an Active straight leg raise:**<br>EMG onset of:<br><br>• Rectus Abdominis<br>• External & Internal obliques<br>• Transversus abdominus | | • Pain<br>• Delayed onset of Transversus abdominis |

Table 2.1: Summary of risk factor research (cont.)

| | | | | |
|---|---|---|---|---|
| **Malliaras et al., 2009** | Case-control | Male Australia football and soccer<br><br>N=29 (10 with groin pain)<br><br>Age: Mean (SD): injured 17.3 (0.8), control 17.1 (1.6)<br><br>Pain for > 4 wks. | • Bent knee fall out ROM<br>• Hip internal rotation ROM<br>• Hip external rotation ROM<br>• Hip adduction strength<br>• Hip abduction strength | • Hip adduction strength |

Table 2.1: Summary of risk factor research (cont.)

| | | | | |
|---|---|---|---|---|
| **Severin et al., 2017** | Case-control | Semi-professional male football<br><br>N=22 (11 with groin pain)<br><br>Age: Mean (range): injured 23 (17 - 28), control 24 (19 - 26)<br><br>Pain: Not reported (History of missed training and/or games in previous 12 months) | **During a kicking task at 45° & 60°:**<br><br>• Pelvic ROM (°) (in all three planes)<br>• Peak Pelvic velocity (°.s$^{-1}$) (in all three planes)<br>• Hip ROM (°) (in all three planes: stance and swing limb)<br>• Peak Hip velocity (°.s$^{-1}$) (in all three planes: stance and swing limb) | Pelvis:<br><br>• Decreased sagittal plane ROM<br>• Decreased transverse plane ROM (45° & 60°)<br>• Decreased rotation velocity (45° & 60°)<br>• Decreased posterior tilt velocity (45° & 60°)<br><br>Hip Swing leg:<br><br>• Decreased sagittal plane ROM (45° & 60°)<br>• Decreased peak flexion velocity (45° & 60°)<br><br>Hip Stance leg:<br><br>• Increase transverse plane ROM (45° & 60°)<br>• Decreased peak flexion velocity (45° & 60°) |

Table 2.1: Summary of risk factor research (cont.)

| | | | | |
|---|---|---|---|---|
| **Edwards et al., 2017** | Case-control | Male Australian football players<br><br>N = 17 (7 with groin pain)<br><br>Age: Mean (SD): injured 17.3 (0.8), control 17.1 (1.6)<br><br>Pain: Not reported | • Age, Height & Body Mass<br>• Bent knee fall out ROM<br>• Hip adductor/abductor ratio<br>• Hip abductor peak force ($Nm.kg^{-1}$)<br>• Hip adductor peak force ($Nm.kg^{-1}$)<br>• Squeeze test (mmHg) at 0, 30 and 45° of hip flexion<br><br>**During a 45° running cut task:**<br>Angles (absolute magnitude & variability)<br><br>At initial contact, peak vertical GRF, weight acceptance*):<br>• T12-T1 angle (in all three planes)<br>• L5- S1 angle (in all three planes)<br>• Hip angle (in all three planes)<br>• Knee angle (in all three planes)<br>• Ankle angle (in all three planes)<br><br>Peak Kinetic (absolute magnitude & variability)<br>• Hip moments (in all three planes)<br>• Knee moments (in all three planes)<br>• Ankle moments (in all three planes)<br>• Vertical GRF at peak & weight acceptance.<br>• Posterior GRF | **Absolute Magnitude**<br>• Decreased hip internal rotation (weight acceptance)<br>• Decreased knee external rotation (weight acceptance)<br>• Increased T12-T1 ipsilateral rotation (initial contact and peak vertical GRF.<br><br>**Variability**<br>• Increased T12-T1 transverse (initial contact and peak vertical GRF)<br>• Increased hip transverse (weight acceptance)<br>• Increased knee transverse (weight acceptance) |

Table 2.1: Summary of risk factor research (cont.)

| | | | | |
|---|---|---|---|---|
| **King et al., 2018** | Cohort Study | Male club level soccer, rugby, GAA football, GAA Hurling and Hockey<br><br>N = 205 (205 with groin pain) (112 completed both the pre and post analysis)<br><br>Age: Mean (SD): injured 24.9 (5.1)<br><br>Pain for > 4 wks. | • HAGOS (all subsections)<br>• Squeeze test (mmHg) at 0, 45 and 90° of hip flexion<br>**During a 110° running cut task:**<br><br>Angles (whole waveform)<br><br>• Thorax (in all three planes)<br>• Thorax relative to pelvis (in all three planes)<br>• Pelvis (in all three planes)<br>• Hip (in all three planes)<br>• Knee (in all three planes)<br>• Ankle (in all three planes)<br>Moments, Powers (whole waveform)<br><br>• Hip (in all three planes)<br>• Knee (in all three planes)<br>• Ankle (in all three planes)<br>Work (as a percentage of total work done at lower limb)<br><br>• Hip (in all three planes)<br>• Knee (in all three planes)<br>• Ankle (in all three planes)<br>• Total reusltant work at Hip, Knee, Ankle | • HAGOS (all subsections)<br>• Squeeze test at 0, 45 and 90° of hip flexion<br>Thorax:<br>• Increased contralateral flexion (0-100%<br>• Increased ipsilateral rotation (0-100%)<br>Thorax on pelvis:<br>• Decreased ipsilateral flexion (16-100%)<br>• Increased ipsilateral rotation (0-90%)<br>Pelvis:<br>• Increased contralateral rotation & tilt (0-100%)<br>Hip:<br>• Decreased flexion (0-100%)<br>• Decrease in hip abduction (67-100%)<br>• Reduced adduction moment (78 -95%)<br>• Decreased extensor moment (50-89%)<br>• Increased sagittal concentric power (68-87%)<br>• Decreased work (total and in all three planes)<br>Knee:<br>• Decreased knee flexion (57-100%)<br>• Increased sagittal concentric power (43- 58%)<br>• Decreased frontal plane and resultant work<br>Ankle:<br>• Increased dorsi-flexion (9-75%)<br>• Increased plantar flexor moment (6 - 71%)<br>• Increased sagittal eccentric power (1 - 24%)<br>• Increased sagittal concentric power (57- 83%)<br>• Increased sagittal plane and resultant work |

Table 2.1: Summary of risk factor research (cont.)

| | | | | |
|---|---|---|---|---|
| **Janse van Rensburg et al., 2017** | Case-control | Male club level soccer, rugby, running and cycling<br><br>N = 20 (10 with groin pain)<br><br>Age: Mean (range):<br><br>Unilateral pain (n = 7): injured 29 (22- 48), control 28.7 (19 - 54)<br><br>Bilateral pain (n = 3): injured 28.7 (27- 39), control 26.3 (20-31) | • Age<br>• Height<br>• Body Mass<br><br>**During a single leg drop landing:**<br><br>Angles at initial contact and lowest vertical position<br><br>• Pelvis angles (in all three planes)<br>• Hip angles (in all three planes)<br>Total ROM<br><br>• Pelvis ROM (in all three planes)<br>• Hip ROM (in all three planes) | Pelvis:<br><br>• Decreased pelvic contralateral tilt (initial contact and lowest vertical position)<br>• Increased pelvis ipsilateral rotation (lowest vertical position)<br>Hip:<br><br>• Decreased hip internal rotation (initial contact)<br>• Increased hip abduction (initial contact)<br>• Increased hip internal rotation ROM |

ROM = Range of motion. GRF = ground reaction force. *Weight acceptance in Edwards et al., (2016) defined as the period between initial contact and the first local minimum of the vertical ground reaction after peak vertical GRF.

### 2.3.7 Variables examined and not identified as risk factors for AGP

In addition to presenting the risk factors for AGP, presenting the variables examined in research but not identified as risk factors for AGP is important to guide future research. This following section will discuss these variables included in the above research.

#### Thorax Mechanics

Accounting for almost 68% of body mass (Winter, 2009) [along with the head and arms], the position of the thorax can have a large influence on the loading experienced by the lower limbs and pelvis (Blackburn and Padua, 2008; Kopper, Ureczky and Tihanyi, 2012; Frank et al., 2013; Sasaki et al., 2015). Only thorax kinematics were investigated in the studies reviewed. While both frontal plane and transverse plane mechanics were associated with AGP (Edwards, Brooke and Cook, 2017; King et al., 2018) [see 3D risk factors for AGP], neither study identified sagittal plane thorax mechanics as being important for AGP.

#### Knee Mechanics

Two studies examined the of effects of AGP on both knee angles and kinetics in all three planes (Edwards, Brooke and Cook, 2017; King et al., 2018). When examining knee kinematics, both the transverse plane and sagittal plane were associated with AGP (Edwards, Brooke and Cook, 2017; King et al., 2018) [see 3D risk factors for AGP] however the frontal plane kinematics was not identified as being related to injury in either study. Similarly with respect to knee kinetics, King et al., (2018) found associations between AGP and frontal and sagittal plane knee kinetics. No research found an association between transverse plane mechanics at the knee and AGP.

### Ankle Mechanics

The ankle joint is important for the modulation of stiffness and load absorption during general dynamic tasks (Farley and Morgenroth, 1999; Lewis and Ferris, 2008; Hobara, Muraoka and Omuro, 2009; Yeow, Lee and Goh, 2011) and may influence mechanics of other proximal segments in the kinetic chain in general. Within this review, two studies examined the effects of AGP on both ankle angles and kinetics in all three planes (Edwards, Brooke and Cook, 2017; King et al., 2018). Despite this, within this review only sagittal plane kinematics and kinetics were associated with AGP (King et al., 2018) [see 3D risk factors for AGP]. Neither study identified ankle transverse or frontal plane mechanics as important for AGP, perhaps due to the dominance that the sagittal plane ankle mechanics has on the above outlined roles of load absorption.

### Age

Age is typically believed to be a non-modifiable risk factor for musculoskeletal injury. As noted by Murphy et al., (2003), this seems reasonable since older players have increased exposure over time within their risk activity. Furthermore, age is typically associated with reduced strength and flexibility, which have been suggested as risk factors for musculoskeletal injury in general. Age was examined as a risk factor for AGP in three studies (Verrall et al., 2007; Edwards, Brooke and Cook, 2017; Janse van Rensburg et al., 2017). Despite this, age was not identified as a risk factor for AGP.

### Height

Height like other anthropometric measurements (mass, body mass index) are commonly examined risk factors for injury. This is because an increase in any one of these factors may increase the absolute force the body must resist. Despite this,

whilst some researchers have associated greater height (Backous et al., 1988) or less height (Orchard, 2001) with injuries in general, other studies have identified no association (Knapik et al., 2001). Three studies examined height as a risk factor for AGP and identified no association (Verrall et al., 2007; Edwards, Brooke and Cook, 2017; Janse van Rensburg et al., 2017)

### 2.3.8 Summary

This chapter section presented studies investigating the risk factors for AGP. Inclusion criteria was similar to that of a previous review (Whittaker et al., 2015) with the exception that only studies investigating groin pain with a defined symptom duration were included in this review. This was to avoid the inclusion of research investigating acute injuries. Given the focus of this thesis on AGP biomechanics, an exception was made for research investigating the 3D biomechanics of groin pain, even if symptom duration was not reported.

Until recently, no research had examined the 3D biomechanical risk factors for AGP. This was unusual given the high prevalence of AGP (O'Connor, 2004; Werner et al., 2009; Murphy et al., 2012) and the fact that biomechanical examinations had been conducted quite extensively with other injuries [e.g. anterior cruciate ligament (ACL) (Lin et al., 2012) and lateral ankle ligament sprain (Willems et al., 2005)]. The four studies presented in this review suggest that the mechanics of the trunk, hip and pelvis may be of importance to AGP. Given the low number of 3D biomechanical studies, it is difficult to identify any clear trends, however the most consistently identified features to be associated with AGP were the pelvis and hip mechanics in all three planes. In particular, the transverse plane appears to be of importance for AGP in these two joints/segments. Two out of two studies identified increased range of motion at the hip in those with AGP in comparison to uninjured controls (Janse van Rensburg et al., 2017; Severin et al.,

2017), while two out of three studies found decreased hip internal rotation angles (Edwards, Brooke and Cook, 2017; Janse van Rensburg et al., 2017). Similarly, at the pelvis two out of three studies found increased pelvis ipsilateral rotation in those with AGP in comparison to uninjured controls (Janse van Rensburg et al., 2017; King et al., 2018). Collectively, these findings suggests that the AGP group are initiating the athletic tasks with greater external rotation of the hip and potentially placing greater loading on the anterior aspects of the hip and pelvis. With the exception of the study by King et al., (2018) however, the sample sizes examined in these studies remain low (n = 7-11), and as such further research is required to substantiate their findings.

For the clinical and demographic measures, the most consistent and strong evidence was for hip adductor weakness (Malliaras et al., 2009; Crow et al., 2010; Nevin and Delahunt, 2014) followed by reduced hip range of motion (Verrall et al., 2007; Nevin and Delahunt, 2014). Whilst altered transversus abdominus function was also identified in two studies (Cowan et al. 2004; J. Jansen et al. 2010), the replication of the findings under artificially induced acute groin pain makes the clinical relevance of transversus abdominus function in AGP questionable (Jansen et al. 2010). Of note was the presence of risk factors bilaterally and not just on the symptomatic side (Morrissey et al., 2012; Nevin and Delahunt, 2014). Whilst this may be due the complex anatomy of the pubic region (Robertson et al., 2009), a predisposition to injury, or a bilateral effect of the AGP, it emphasises the need to look beyond asymmetry in AGP and highlights the perils of using the non-symptomatic side as a reference in clinical examinations.

As previously noted there remains a paucity of research investigating the 3D biomechanics of AGP, typically utilising small sample sizes. Future research is therefore required to increase the breath and quality of research investigating the Biomechanical risk factors for AGP. In addition to traditional examinations of

kinematics and kinetics, the affects of AGP on stiffness and movement variability warrent investigation.

## 2.4 Stiffness

There has been growing interest in the association stiffness has with respect to injury (Butler, Crowell and Davis, 2003; Brazier et al., 2014). Despite this, while the relationship between stiffness and performance is well established (McMahon, Comfort and Pearson, 2012; Brazier et al., 2014), literature investigating specific injuries and stiffness remains limited. In particular, to date no research has investigated if stiffness is affected by AGP (see section 2.3).

### 2.4.1 Defining and Calculating Stiffness

Stiffness as a mechanical measure is well defined (Latash and Zatsiorsky, 1993), however its calculation within human tissues has led to some inconsistencies (Serpell et al., 2012). The following section will address both the definition of stiffness and common forms of its calculation within human movement research. As noted, the mechanical or physics definition of stiffness is well defined and has its origin with Hooke's Law:

$$F = k.x \tag{2.1}$$

Hooke's Law states that the force (F) required to deform a spring is proportional to a constant (k) and the distance (x) the spring is deformed, provided that its shape is not permanently changed (Latash and Zatsiorsky, 1993, Butler et al., 2003). The constant term k in this equation is the stiffness of the deformable spring, which under the influence of external forces can store elastic energy (Latash and Zatsiorsky, 1993). Whilst the spring-like behaviour of muscles and joints has

been known since 1847 (Weber, 1847 cited Latash and Zatsiorsky, 1993), the strict physics definition of stiffness cannot be applied to the human body. According to Butler et al. (2003) an ideal spring is massless, moves in one direction only, and has a stiffness that is independent of time, length, or velocity. Furthermore, Latash and Zatsiorsky, (1993) suggest that an accurate stiffness model must account for all physical characteristics of the body and be able to describe changes in muscle force as a function of contraction velocity. Given the complexity of such a model, Latash and Zatsiorsky, (1993) proposed the use of a simplified 'quasi-stiffness', whereby requirements for the system being at equilibrium and the time course of displacement can both be disregarded. While it is important to note the distinction between true mechanical stiffness and quasi-stiffness, for simplicity the term 'stiffness' will be utilised to represent quasi-stiffness for the remainder of this thesis.

### 2.4.2 Stiffness in humans

Stiffness in humans has been examined at various levels of the body, from a cellular level (Wells, 1981) to a whole body measure (Blickhan, 1989; McMahon and Cheng, 1990). However, the most commonly examined forms of stiffness with respect to injury are whole body and joint stiffness and will as such form the focus of this review.

#### Whole body stiffness

At its most simple form, the human body can be modelled as spring mass model (Figure 2.4).

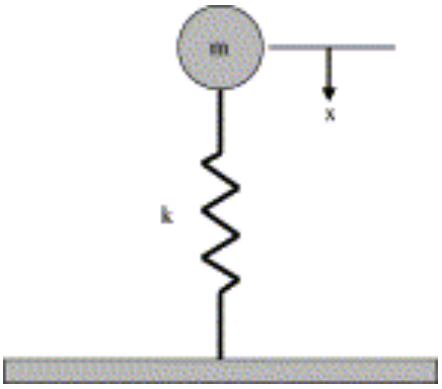


Figure 2.4: Ideal spring and mass model used for calculating vertical stiffness (Butler, Crowell and Davis, 2003)

According to this model, the legs act like a spring during running gait to firstly absorb forces during landing and then subsequently reutilise this energy during the take-off phase. When non-vertical displacements are negligible (e.g. hopping on the spot) stiffness can be calculated as vertical stiffness and represents, at a whole body level, the resistance of the centre of mass to vertical displacement under a given vertical ground reaction force (McMahon and Cheng, 1990):

$$Vertical\ stiffness = \Delta Fz/\Delta COMz \tag{2.2}$$

Where $\Delta$ Fz is the change in vertical ground reaction force and $\Delta$ COMz is the vertical displacement of the centre of mass. When non-vertical displacements are not negligible (e.g. running) whole body stiffness is more appropriately modelled as leg stiffness (McMahon and Cheng, 1990):

$$Leg\ stiffness = L_0/mg = (Fz/mg)/[1 - cos\theta + (\Delta COMz/L_0)] \tag{2.3}$$

Where mg is the subject's weight, $L_0$ is the rested state leg length and $\theta$ is the angle of attack (defined as half the angle swept by the leg). Both vertical stiffness and leg stiffness are equivocal during vertical hopping as the angle of attack is 0. However, during tasks such as running, leg stiffness will always be less than

vertical stiffness given the inclusion of the angle of attack as a denominator in the equation.

**Joint stiffness**

A similar concept to the above linear measures of stiffness is joint stiffness. While the lower limbs are commonly modelled as a single spring, the actual stiffness of the leg spring depends on adjustments to the joint angles of the lower extremity. Joint stiffness is calculated as the ratio between change in angle and applied torsional force and reflects, similar to vertical stiffness, the resistance of a joint to rotation under a given moment of force (Farley et al., 1998).

$$Joint\ stiffness = \Delta M / \Delta \theta \tag{2.4}$$

Where $\Delta$M is the change in joint moment and $\Delta\theta$ is the range of motion of the joint.

### 2.4.3 Stiffness and Injury

Despite, the growing interest in stiffness and its relationship to injury (Brazier et al., 2014), there remains a paucity of studies directly investigating this topic. It has been suggested that inappropriate levels of vertical and/or joint stiffness, either insufficient or excessive, may elevate the risk of injury (Butler, Crowell and Davis, 2003). On the one hand high levels of stiffness may result in greater magnitudes of peak loading when compared to low levels of stiffness (Milner et al., 2006). As per the impulse momentum equation (Equation 2.5), for a given impulse of loading, less joint excursion will result in higher peak and higher rates of loading in ground reaction forces, as the time over which the body is brought to rest is reduced.

$$F * t = m * \Delta v \tag{2.5}$$

Where on the left hand side of the equation, F is force; t is time (and cumulatively represents impulse). On the right hand side of the equation, which represents the change in momentum of the body, m is mass and $\Delta$v is the change in the body's velocity. While, the theoretical basis for the relationship between high levels of stiffness and injury is clear, it has also been suggested that insufficient stiffness may also increase the risk of overuse injury. During landing, low levels of stiffness may increase the range through which a joints will travel, thus potentially increasing localised loading on the tissues surrounding a joint (Maquirriain, 2012). However, clearly the possibility for increased localised loading will depend on the type of joint and if the muscles surrounding the joint support or load connective tissue.

Early investigations into stiffness and injury were in relation to foot structure in runners (Williams et al., 2004). The authors identified that in uninjured populations, high arched runners had increased leg stiffness and vertical loading rates compared with low-arched runners. The authors suggested that the differing levels of leg stiffness were likely responsible for the pattern of injuries observed in an earlier study (Williams, McClay and Hamill, 2001), whereby high arched runners had greater incidence of bony injuries compared with low-arched runners, while low arch runners were more likely to develop soft tissue injuries. In support of this concept, Maquirriain (2012) evaluated leg stiffness in patients suffering unilateral Achilles tendinopathy. The authors identified that leg stiffness was significantly reduced in patients with unilateral Achilles tendinopathy. Milner et al., (2006) investigated sagittal plane knee and ankle joint stiffness in male and female runners with a history of tibial stress fracture in comparison to uninjured runners. The authors identified no significant difference between the

two groups, but did demonstrate a trend (p = 0.054) with moderate effect size (d = 0.54) towards higher stiffness at the knee joint for the stress fracture group in comparison to the uninjured group. Later research by Milner, Hamill and Davis, (2007), reinvestigated this using a more homogeneous group of female athletes. The authors identified that sagittal plane knee stiffness during the initial loading of stance was significantly greater in the stress fracture group compared with the uninjured group. Despite the growing interest into vertical, leg and joint stiffness and their relation to injury, there remain too few studies investigating this topic to allow for formative conclusions. Current research tends to support the notion that there is an optimal range of stiffness outside of which, the risk of injury increases (Butler et al. 2003). However, to date no research has investigated stiffness in relation to AGP as proposed in this thesis.

### 2.4.4 Summary

It has been suggested that improper modulation of stiffness during athletic tasks may increase the risk of injury. While there are numerous forms of stiffness measured in humans, the most commonly examined forms with respect to injury are joint and whole body stiffness. While some authors have identified that low levels of stiffness may increase the risk of certain injuries, other authors have identified an associative relationship between high stiffness and injury. With respect to AGP, stiffness may be of importance to the development of the condition as any alteration in the magnitude of loading or the manner in which loads are absorbed may overload the musculotendinous and bony structures surrounding the pubic symphysis region (Meyers, Greenleaf and Saad, 2005; Franklyn-Miller et al., 2017). To date however no research has examined stiffness in AGP patients as proposed in this thesis.

## 2.5 Variability and Complexity

With traditional forms of biomechanical analysis (e.g. measurements of kinematics/ kinetics and measures of stiffness) variability is treated as an error in movement or noise arising from technical or measurement sources (Racic, Pavic and Brownjohn, 2009; Taylor et al., 2015; Gore et al., 2016). Recently however, intra-individual variation in movement patterns have been considered an integral characteristic of any motor task (König et al., 2016) allowing flexible adaptations to stresses placed on the human body (Stergiou, Harbourne and Cavanaugh, 2006). As such it is thought that movement variability may have a functional role to play with respect to musculoskeletal injury (Bartlett, Wheat and Robins, 2007; Stergiou and Decker, 2011; Hamill, Palmer and Van Emmerik, 2012) and provide additional information not captured in more traditional biomechanical measures [e.g. stiffness (see section 2.4)]

Movement variability can be assessed in terms of both the amount of variability and the structure of variability, which functionally represent different aspects of movement variability (Newell and Slifkin, 1998) and can vary independently of one another (Harbourne and Stergiou, 2009). To quantify the amount of movement variability, linear statistical tools are utilized (e.g. standard deviation, coefficient of variation), whereas to quantify the structure of variability non-linear tools are utilized [e.g. sample entropy, Lyapunov exponent] (Harbourne and Stergiou, 2009). Within this review of literature, the amount of linear variability will be examined systematically and has been published as such (Baida et al., 2017). While it would be of value to systematically review the literature on both forms of analysis, there are currently far fewer studies on the structure of variability, preventing robust conclusions from being drawn. Herein the term variability will be used to refer the magnitude of linear variability. In contrast when referring to

the structure of variability or non - linear variability the term complexity will be used.

### 2.5.1 Complexity - a non linear variability

Complexity is a hallmark of physiological systems, both in structure and function (Goldberger et al., 2002). While no consensus definition of complexity exists, it is accepted that the complexity of a signal relates to its structural richness and correlations across multiple time scales (Goldberger, Peng and Lipsitz, 2002). Furthermore, complex signals are characterised by being non-stationary (i.e. they are not stable) and non-linear in nature. The term non-linear applies to systems whose components interact in a non-additive way (Goldberger, Peng and Lipsitz, 2002). In contrast to linear systems, non-linear systems are characterised by a lack of proportionality, with small changes having possibly striking, unpredictable effects, thus limiting the ability to predict their long-term behaviour (Stergiou and Decker, 2011).

While these complex signals were traditionally seen as noise in the system, it has been acknowledged that these fluctuations contain meaningful structural richness (van Emmerik 2002). Further, it is now accepted that healthy organisms are defined by high level of non-linear variability or complexity, that is, they produce complex signals (Stergiou, 2016). This complexity arises from the functional interaction of numerous system elements and regulatory feedback loops (van Emmerik 2002). A complex system is considered healthy as it reflects the system's capacity to respond to a constantly changing environment (Goldberger et al., 2002; Lipsitz, 2002). Pathology or injury on the other hand results in a reduction in the number of and/or communication between systems of the body, ultimately reducing system complexity (Lipsitz, 2002).

As noted by Goldberger et al., (2002), the non-stationary and non-linearity nature of physiological signals generated by living organisms is not well quantified by traditional biostatistical methodologies. As such, this has led to the development of a wide range of statistical and mathematical techniques to help understand and quantify some of this 'hidden information' that physiological signals contain. There are a number of techniques that can be utilised to examine complexity, for example; 'entropy', 'detrended fluctuation analysis' and 'lyapunov exponent'. While it is acknowledged that reliance on any single test may give a misleading representation of physiological complexity (Goldberger 2002), a review of all available methods is outside the scope of this current thesis. Furthermore, given that complexity was not the sole focus of this thesis, it was decided to focus on only one measure to explore the utility of complexity in this area. This current chapter section will focus on one of the most useful and frequently applied of these methods, namely, entropy, which was utilised within this thesis. Readers interested in other forms of complexity are referred to the following text (Stergiou, 2016).

**Entropy**

The concept of entropy was developed in classical thermodynamics, where it grew out of the work on steam engines [Carnot (1824) cited (Stergiou, 2016)]. However, its modern implementation comes from information theory, where entropy is defined as the loss of information in a time series or signal (Richman and Moorman, 2000). In this respect, if a system has very low entropy, the next state of the system is very predictable, that is, it has high regularity. Within the biological literature, a reduction in entropy has typically been associated with pathology and a reduction in complexity. For example both Cavanaugh et al., (2006) and Quatman-Yates et al., (2015) found reduced balance entropy in players post concussion in comparison to pre-season measures and uninjured

controls, respectively. Interestingly, despite traditional linear measures of balance normalizing, entropy remained depressed in those who had suffered a concussion suggesting that complexity may be a more sensitive measure underlying injuries than traditional linear measures (Cavanaugh et al., 2006). In the area of cardiology, entropy has been widely used to detect various pathologies including ventricular dysfunction (Fleisher, Pincus and Rosenbaum, 1993) and as early indicator of atrial fibrillation (Alcaraz and Rieta, 2010). Entropy has also been used to determine that children with cerebral palsy have a more regular postural sway than typical children (Donker et al., 2008) and as an indicator of reduced functional capacity associated with ageing (Kaplan et al., 1991) and detraining (Heffernan et al., 2007). Indeed this association between pathology and entropy has led to authors utilizing the phrase 'loss of complexity hypothesis' in relation to reduced entropy (Stergiou, 2016). It is noted however, that pathology may exist at both the ends of the entropy spectrum (Stergiou and Decker, 2011) and that a decrease in entropy does not necessitate a decrease in complexity (Goldberger, Peng and Lipsitz, 2002). While acknowledging this, given the close relationship between entropy, pathology and (by association) complexity within the scope of human movement, for the remainder of this thesis a loss of complexity will be used synonymously with reduced entropy. Despite its widespread use in general biological literature and motor control, there remains a dearth of studies investigating the affect of musculoskeletal injury/pain on movement complexity using entropy (Table 2.2).

Table 2.2: Studies investigating entropy and musculoskeletal injury/pain

| Authors | Task | Subjects | Study Design | Variables | Entropy Metric | Results |
|---|---|---|---|---|---|---|
| **Georgoulis et al., (2006)** | Walking on treadmill | 10 subjects (8 males, 2 females) with ACL rupture<br><br>Age: 34.7 ± 11.1 | 2 min walks at 120% and 80% of comfortable speed; ~80 strides collected at each speed | Flexion/extension (sagittal) knee angular displacements | Approximate entropy | Injured knee < Uninjured Knee |
| **Moraiti et al., (2009)** | Walking on treadmill | 12 male subjects post ACL reconstruction<br><br>[6 patellar graft (PT) and 6 semitendinosus graft (ST)] and 6 male controls<br><br>PT Age: 24 ± 3<br><br>ST Age: 27 ± 6<br><br>Control Age: 28 ± 3 | 2 min at self selected pace | Flexion/extension (sagittal) knee angular displacements | Approximate entropy | ACL (PT & ST) > Control |
| **Tochigi et al., (2012)** | Over ground indoor walk | 52 subjects with knee osteoarthritis ( 17 male 35 female) and 57 controls (27 male , 30 female)<br><br>Injured Age: 22.48 ± 3.98<br><br>Control Age: 21.56 ± 3.15 | 400m walk (10 * 20m shuttle) at self selected fastest pace possible | Tibia resultant accelerations | Sample Entropy | OA < Controls<br><br>Injured knee < Uninjured Knee |

Table 2.2: Studies investigating entropy and musculoskeletal injury/pain (cont.)

| | | | | | | |
|---|---|---|---|---|---|---|
| **Søndergaard et al., (2010)** | Prolonged sitting | 9 healthy Males<br><br>Age: 25.2 ±1.6 | ~90 min sitting | centre of pressure and lumbar curvature | Sample entropy | Pain < No Pain |
| **Alkjaer et al., (2015)** | Walking on treadmill | 11 females with knee osteoarthritis and 11 controls<br>Injured Age: 69.0 ± 6.6<br>Control Age: 66.1 ± 4.5 | 3.5 km/h walking speed | Stride intervals | Sample entropy | Controls = OA |
| **Terada et al., (2015)** | Walking on treadmill | 25 with self reported chronic ankle instability (14 male & 11 female) and 27 controls (10 male & 17 female)<br>Injured Age: 54 - 79<br>Control Age: 21 - 79 | 3 min walking at self selected speed | Lower extremity kinematics in the sagittal and frontal planes | Sample entropy | Injured < Control<br>(Ankle frontal plane) |
| **Schütte et al., (2017)** | Fatiguing outdoor run (3200m) | 14 with medial tibial stress syndrome (8 male, 6 female) and 16 controls (10 male, 6 female)<br>Injured Age: 20.36 ± 0.84<br>Control Age: 20.13 ± 0.72 | Maximal effort run | Trunk IMU accelerations in all three planes | Sample entropy | Injured < Control with fatigue<br>(medio lateral accelerations) |

ACL = Anterior Cruciate Ligament, OA = Osteoarthritis

Of the seven studies investigating entropy and musculoskeletal injury or pain, five studies (72%) found lower entropy in injured/painful subjects in comparison to controls and/or the controlled setting (Georgoulis et al., 2006; Söndergaard et al., 2010; Tochigi et al., 2012; Terada et al., 2015; Schutte et al., 2018). In comparison, one study (14%) found greater entropy in the injured group in comparison to uninjured controls (Moraiti et al., 2009) and one study (14%) found no difference between injured and uninjured controls for measures of entropy (Alkjaer et al., 2015). These results demonstrate a trend towards reduced complexity in the injured group in comparison to the uninjured group. The most popular form of entropy utilized in the musculoskeletal injury literature is sample entropy, which accounted for five out seven of the studies reviewed (72%). To date no research has investigated the affect of AGP on complexity and/or entropy. Clearly given the importance of this measurement with respect to human pathology, complexity may also be a useful measure to delineate between those with and without AGP.

### 2.5.2 The magnitude of linear variability

This review of literature has previously been published in full:

Baida, S.R., Gore, S.J., Franklyn-Miller, A.D. and Moran, K.A. 2017. Does the amount of lower extremity movement variability differ between injured and uninjured populations? A systematic review. Scandinavian Journal of Medicine & Science in Sports.

It is presented here in this chapter with only minor changes to conform to the style and formatting of this thesis.

It has been postulated within the framework of dynamic systems theory (Hamill et al., 1999), that reduced variability during movement might lead to

repetitive loading on a specific tissue structure resulting in excessive stress and eventual injury (Harbourne and Stergiou, 2009). Several authors have provided further support for the association between injury and reduced variability in various injured groups including chronic ankle instability (Herb et al., 2014) and patellar tendinopathy (Kulig, Joiner and Chang, 2015). However, in direct contrast it has also been suggested that greater variability is associated with injury (Stergiou, Harbourne and Cavanaugh, 2006; Hamill, Palmer and Van Emmerik, 2012). In accordance with general motor program theory variations in movement may represent aberrant neuromuscular motor control (Schmidt, 2013) resulting in poorly controlled actions which may lead to excessive stress and injury (Hamill, Palmer and Van Emmerik, 2012). In support of this theory, numerous studies have found greater variability in a number of different injury groups including athletic groin pain(Edwards, Brooke and Cook, 2017), chronic ankle instability (Kipp and Palmieri-Smith, 2012) and iliotibial band syndrome(Miller et al., 2008). With such contrasting evidence in the literature, there is a clear need for a systematic review investigating movement variability and its relationship with injury. To date this has not been conducted.

Both discrete (e.g. peak knee moment) and continuous measures (e.g. 0-100% of the knee moment waveform) have been examined when investigating inter-trial variability. Given that analysis of the whole continuous waveform has been shown to be more effective at detecting differences between groups (Richter, O'Connor, et al. 2014c; Marshall et al. 2015) it would be useful to explore whether findings on movement variability is also affected by the type (discrete versus continuous) of analysis employed. The primary aim of this systematic review is to investigate published comparison trials to determine if the amount of movement variability in dynamic tasks differs between groups with lower limb musculoskeletal injury and uninjured controls. In light of the diverse range of

Table 2.3: Search terms

| | |
|---|---|
| Population | injur* OR musculoskeletal* |
| Outcome | Variability |
| Variables | Biomech* OR kinetic OR kinematic OR motor control OR coordination OR dynamic systems |
| NOT | upper* OR spine OR lumbar* OR arm OR back OR heart rate OR animal* OR Cadaver* OR rat* OR monkey OR frog OR robot* OR modelling OR pharma* OR mice OR cat* OR fish* OR DNA OR gene OR RNA |

methods used, the secondary aim of this review is to provide methodological recommendations for future research.

### Methodology

**Protocol and Registration** This systematic review was registered (42016039113) with Prospero (Centre for Reviews and Dissemination), University of York, on 13/5/2016. The Preferred Reporting Items for Systematic reviews and Meta-Analyses (PRISMA) statement was used (Liberati et al., 2009).

**Search Strategy** A systematic search of the literature was undertaken independently by two authors (SB, SG) between June and September 2016 for clinical trials, case-comparison and cohort studies comparing movement variability between injured populations and uninjured controls. Databases, searched from inception until 01/09/16, included Medline, Sports Discus, Scopus and Web of Science. The terms applied in all database searches are presented in Table 2.3. These were combined using relevant Boolean terms and limits of English language and human population were placed on searches.

**Selection Criteria** Three authors (SB, SG, KM) determined the selection criteria, as listed in Table 2.4, before commencing the search. This review only addresses measures that assess the *amount* of variability in movement, determined by linear analysis techniques applied to a time series (e.g. standard deviation, coefficient of

Table 2.4: Inclusion and exclusion for literature search

| | |
|---|---|
| • Included participants with a lower limb injury<br>• Compared injured participants to uninjured controls,<br>• Examined movement variability for at least one dependent variable,<br>• Provided a statistical between-group comparison when comparing measures of movement variability | • Investigated neurological disorders<br>• Examined musculoskeletal injury in the upper extremity or spine<br>• Used non-linear measures to examine variability (i.e. complexity) |

variation, linear variation, range and circular variants of these metrics when examining coordination variability). This review does not examine non-linear analysis techniques (e.g. entropy measures and lyapunov exponent) that investigate the structure of variability (complexity) throughout a time series.

All studies investigating the amount of movement variability in lower extremity musculoskeletal injuries were included in this review. Musculoskeletal injury was defined as any acute or chronic injury episode that would have influenced both the peripheral (tissue damage) and central (spinal cord and brain) movement systems that may have led to altered movement patterns and variability of movement patterns. While it may be argued that patients who have undergone anterior cruciate ligament reconstruction (ACLR) are in fact recovered, we have included this population in this review. This is justified as potentially patho-mechanical movement strategies remain evident in this population 6 or more months following surgery (Gokeler, Hof and Arnold, 2010). The two authors (SB, SG) independently applied the selection criteria when reviewing titles and abstracts and a full review of a manuscript was performed if selection was unclear. Disagreements were resolved by discussion or third party consultation (KM). Reference lists of selected studies were examined in order to identify further relevant studies.

**Data extraction and analysis** Data extraction was performed independently by the authors (SB, SG) using predefined data fields and then crossed-checked for accuracy. Data identified for qualitative analysis included type and location of musculoskeletal injuries, analysis techniques implemented, dependent variables utilised and physical tasks performed. The principle quantitative measure extracted for this review was the probability value used to identify significant between-group differences within a study. All of the variables examined within this review are listed within Appendix A along with the significant findings reported in the related studies. A meta-analysis of the studies was not possible due to the limited reporting of values and data for any one biomechanical measure. A qualitative analysis was therefore implemented to provide a best evidence synthesis and where possible a subgroup analysis between studies was performed (Pietrosimone et al., 2008; Santamaria and Webster, 2010; Fong Yan et al., 2013; Aderem and Louw, 2015; Undheim et al., 2015). Where one study utilised various methodological approaches (tasks examined included both run and walk conditions, the dependent variables examined included both kinematic and spatiotemporal, or the analysis technique included both continuous point-by-point and discrete-point measures), both components were considered separately for analysis in the results section of this review.

**Assessment for risk of bias** Assessment for risk of bias A modified version of the Downs and Black's checklist (Downs and Black, 1998) as proposed by Trac et al. (Trac et al., 2016) was used to assess study quality (subscales include; reporting, external validity, internal validity, selection bias and power). This version changed the scoring of question 27 from 5 to 2, making 29 the total score of the 27 questions (Trac et al., 2016). Studies were appraised independently (SB, SG).

## Results

**Overview of findings** The search method employed in this review identified 1053 studies. After titles and abstracts were reviewed 69 studies were retrieved; duplicates were then removed, leaving 37 studies for a full review. Of these 37 studies, 20 were excluded for the following reasons: (1) investigated variability using nonlinear approaches only (measures of complexity) (n=7), (2) provided no control group (n=5), (3) provided no inferential statistical comparison of movement variability (n=3), (4) did not measure variability (n=2), (5) were review papers (n=2), or (6) not relevant (n=1). This left 17 papers included for review from the database searches. Pearling of article bibliographies adhering to the inclusion criteria, revealed an additional five studies adhering to the inclusion criteria. A final 22 papers were incorporated in this systematic review, which included the findings from 295 injured subjects and 319 uninjured controls. Figure 2.5 presents a flow diagram of study inclusion.

Quality appraisal scores using the modified Downs and Black's checklist (Trac et al., 2016), presented in full in appendix A, ranged from 11 to 14 out of a possible 29 points with a median score of 12. No study received a point on the criterion scoring external validity (question 11, 12 and 13). Also criterion not fulfilled by any study included questions related to blinding, follow-up of patients' post-intervention and compliance with the intervention (questions 14, 15, 17, 19). Discrepancies in scores were based on points attained under the 'reporting' and 'power' sub-scales. Eighteen studies (Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Ferber et al., 2005; Miller et al., 2008; Drewes et al., 2009; Chiu, Lu and Chou, 2010; Meardon, Hamill and Derrick, 2011; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Hein et al., 2012; Herb et al., 2014; Cunningham et al., 2014; Kulig, Joiner and Chang, 2015; Mann et al., 2015; Paquette, Milner and Melcher, 2016; Gribbin et al., 2016; Hamacher, Hollander and

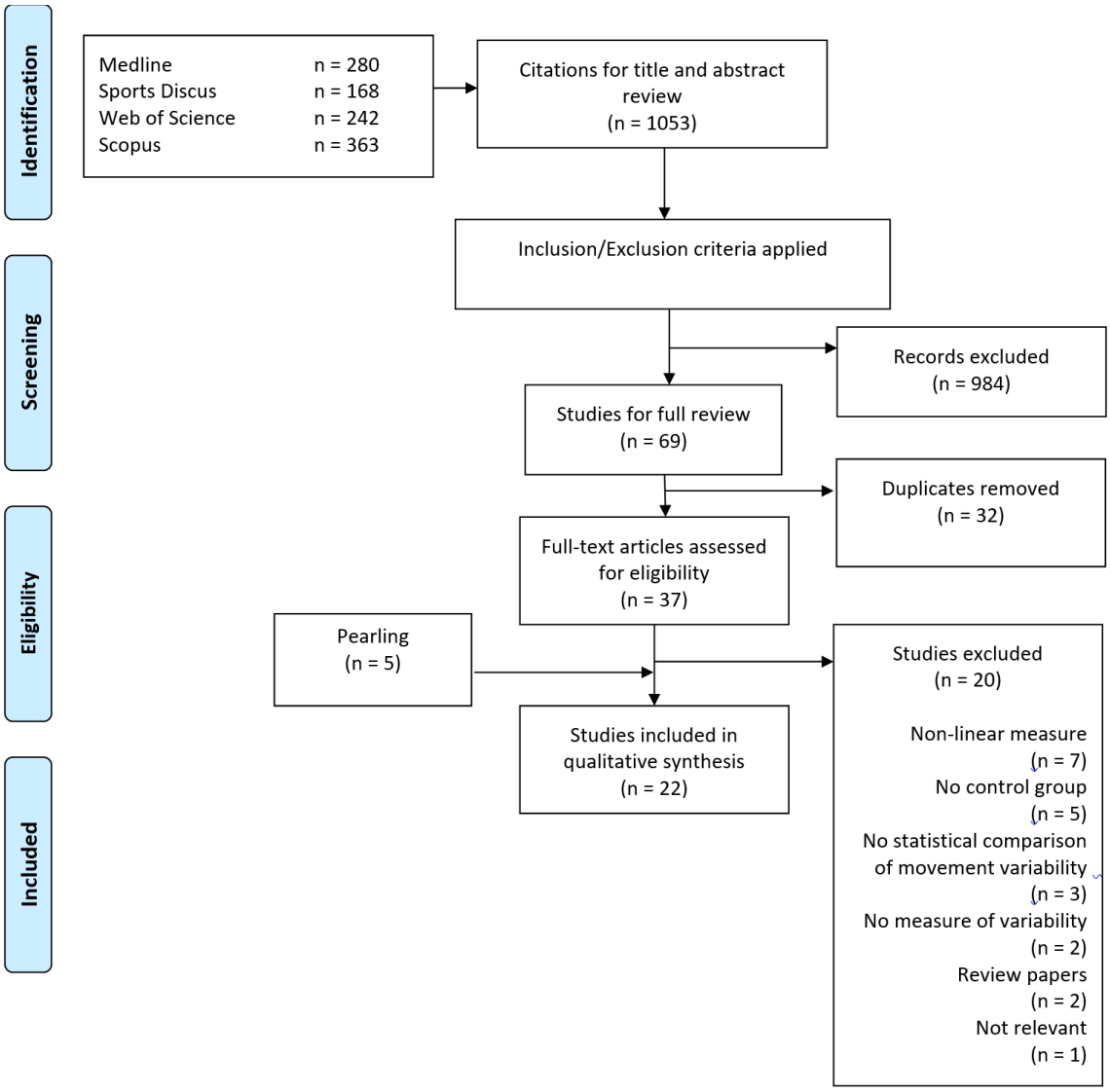


Figure 2.5: PRISMA flow diagram of search strategy

Zech, 2016; Edwards, Brooke and Cook, 2017) scored a point for clearly describing included subjects (criterion 3). Four studies (Heiderscheit, Hamill and Emmerik, 2002; Miller et al., 2008; Cunningham et al., 2014; Edwards, Brooke and Cook, 2017) scored a point for reporting adverse outcomes resulting from performance of the task (criterion 8). Eighteen studies (Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Ferber et al., 2005; Miller et al., 2008; Chiu, Lu and Chou, 2010; Maclean et al., 2010; Meardon, Hamill and Derrick, 2011; Brown, Bowser and Simpson, 2012; Hein et al., 2012; Kipp and Palmieri-Smith, 2012; Cunningham et al., 2014; Kulig, Joiner and Chang, 2015; Mann et al., 2015; Pollard

et al., 2015; Cordeiro et al., 2015; Paquette, Milner and Melcher, 2016; Hamacher, Hollander and Zech, 2016; Edwards, Brooke and Cook, 2017) scored a point by reporting actual probability values (criterion 10). Ten studies (Ferber et al., 2005; Miller et al., 2008; Drewes et al., 2009; Maclean et al., 2010; Brown, Bowser and Simpson, 2012; Hein et al., 2012; Cordeiro et al., 2015; Mann et al., 2015; Paquette, Milner and Melcher, 2016; Edwards, Brooke and Cook, 2017) scored a point for reporting calculation of sample size (criterion 27).

A summary of the main subject and analysis characteristics extracted for each study is presented in Table A.2. Overall findings revealed that 73% (n=16/22) of studies reported a statistically significant difference in at least one dependent variable used to examine movement variability between injured subjects and uninjured controls. Injured subject groups demonstrated greater variability in 64% (n=14/22) of the studies (James, Dufek and Bates, 2000; Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Miller et al., 2008; Chiu, Lu and Chou, 2010; Maclean et al., 2010; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Cunningham et al., 2014; Pollard et al., 2015; Cordeiro et al., 2015; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016; Edwards, Brooke and Cook, 2017), reduced variability in 27% (n=6/22) (James, Dufek and Bates, 2000; Miller et al., 2008; Brown, Bowser and Simpson, 2012; Herb et al., 2014; Kulig, Joiner and Chang, 2015; Gribbin et al., 2016) and no difference between groups was evident in 27% (n=6/22) (Ferber et al., 2005; Drewes et al., 2009; Meardon, Hamill and Derrick, 2011; Hein et al., 2012; Mann et al., 2015; Paquette, Milner and Melcher, 2016). Table 2.6 presents the percentage of studies reporting greater, less or no difference in variability when comparing injured subjects to uninjured controls.

Table 2.5: Summary of subject and analysis characteristics

| Author | Subject characteristics | | | | | | Analysis characteristics | | | | Significant Finding |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | Group | Injured | Uninjured | Impaired region | Variability Analysed | Pain Present | Task Performed | Number of Trials | Additional Technique | Variability Metric | |
| **ACLR** | | | | | | | | | | | |
| Van Uden et al. 2003 | ACLR | 13 | 7 | knee | Knee- ankle | No | Single leg hopping | 10 seconds | CRP | SD | **> Knee-Ankle** |
| Cordeiro et al. 2015 | ACLR | 8 | 9 | Knee | Knee | Yes | In-step soccer kick | 3 kicks | - | SD | **> Knee** |
| Pollard et al. 2015 | ACLR | 10 | 10 | Knee | Hip-knee | NR | 45° side-step cut | 4 cut trials | VC | SD | **> Hip-Knee** |
| Gribbin et al. 2016 | ACLR | 22 | 15 | Knee | Hip-knee | NR | Run/walk | 10 strides | VC | VCV | **> Hip-Knee**<br>**< Hip-Knee** |
| **Ligamentous** | | | | | | | | | | | |
| Kipp and Palmieri-Smith 2012 | CAI | 11 | 11 | Ankle | Ankle | Yes | Single leg land | 5 trials | FPCA | SD | **> Ankle** |
| Brown, Bowser, and Simpson 2012 | CAI | 46 | 24 | Ankle | Trunk, hip, knee, ankle | No | 2 leg jump – single leg land (3 directions) | 10 trials each condition | - | CV | **> Trunk**<br>**< Hip, Knee** |
| Hamacher, Hollander, and Zech 2016 | CAI | 12 | 12 | Ankle | Ankle | NR | Run | 40 strides | - | SD | **> Ankle** |
| Drewes et al. 2009 | CAI | 7 | 7 | Ankle | Shank, rearfoot | No | Run/Walk | 21 run strides 14 walk strides | CRP | DP | **=** |
| Herb et al. 2014 | CAI | 13 | 15 | Ankle | Shank, rearfoot | NR | Run/Walk | 3 strides | VC | VCV | **< Ankle** |
| **Tendon** | | | | | | | | | | | |
| Kulig, Joiner, and Chang 2015 | PT | 9 | 9 | Knee | Hip, knee, ankle | No | Land (from jump spike) | 3 -6 trials | - | CV | **< Ankle** |
| **Overuse Injury** | | | | | | | | | | | |
| James, Dufek, and Bates 2000 | Injury Prone | 10 | 10 | Lower limb | Hip, knee, ankle | No | Land | 10 each condition | - | SD | **> Ankle**<br>**< Ankle** |

Table 2.5: Summary of subject and analysis characteristics (cont.)

| | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Ferber, Davis, and Williams 2005 | RRI | 11 | 11 | Lower limb | Tibia-rearfoot | No | Run | 5 trials | VC | SD | = |
| Mann et al. 2015 | RRI | 44 | 46 | Lower limb | Spatiotemporal | No | Run | 161 strides | - | CV | = |
| Paquette, Milner, and Melcher 2016 | RRI | 23 | 21 | Lower limb | Ankle | No | Run | 5 foot strikes @ 10 min intervals | - | SD | = |
| Maclean, Emmerik, and Hamill 2010 | RRI | 9 | 9 | Knee | Tibia-calcaneus, knee-rearfoot | No | Run | 10 seconds In each condition | VC | SD | **> Tibia-calcaneus** |
| Meardon, Hamill, and Derrick 2011 | RRI | 9 | 9 | Lower limb | Temporal | No | Run | 661 strides | - | SD, CV | = |
| Miller et al. 2008 | ITBS | 8 | 8 | Knee | Thigh-tibia, thigh-foot, tibia-foot, knee-foot | No | Run | 10 seconds @ 2min intervals | CRP | SD | **> Knee-Foot < Thigh-Foot, Tibia-Foot** |
| Hein et al. 2012 | ITBS | 18 | 18 | Knee | Hip-knee, knee-ankle, | Yes | Run | 5 stance phases | CRP | SD | = |
| Heiderscheit, Hamill, and Emmerik 2002 | PFPS | 8 | 8 | Knee | Thigh-leg, knee-ankle | Yes | Run | 15 strides | VC | CV | **> stride length** |
| Cunningham et al. 2014 | PFPS | 11 | 19 | Knee | Knee-ankle | Yes | Run | 5 strides | VC | Mean, SD | **> Knee-Ankle** |
| Edwards et al. 2016 | AGP | 10 | 7 | Hip | Ankle, knee, hip, trunk | No | Side step cut | 10 cut trials | - | CV | **> Trunk** |
| **Osteoarthritis** | | | | | | | | | | | |
| Chiu, Lu, and Chou 2010 | THA | 20 | 10 | Hip | Hip-knee, knee-ankle | Yes | Walk | 10 meters | CRP | SD | **> Hip-Knee, Knee-Ankle** |
| **Mean / Median** | | | | | | | | **Mean / Median** | | | |
| | | 15 / 11 | 13 / 10 | | | | **Strides** | 85 / 14 | | | |
| | | | | | | | **Trials** | 6 / 5 | | | |

ACLR - anterior cruciate ligament reconstruction, CAI - chronic ankle instability, PT - patellar tendinopathy, RRI - running related injury, ITBS - iliotibial band syndrome, PFP - patellofemoral pain, AGP - athletic groin pain, THA - total hip arthroplasty. NR - not reported. CRP - continuous relative phase, VC - vector coding, VCV - vector coding variability, FPCA - functional principal component analysis, SD - standard deviation, CV coefficient of variation, DP - deviation phase

### Overview of findings

**Findings by Injury Type** A wide variety of different injury types were identified within the studies. Subject groups consisted of either individuals with a single specific injury or various lower limb injuries. In the studies investigating single specific injuries, 88% (n=14/16) reported significant between-group differences. Of these, greater variability was evident in 75% (n=12/16) of them (Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Miller et al., 2008; Chiu, Lu and Chou, 2010; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Cunningham et al., 2014; Pollard et al., 2015; Cordeiro et al., 2015; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016; Edwards, Brooke and Cook, 2017), reduced variability in 31% (n=5/16) (Miller et al., 2008; Brown, Bowser and Simpson, 2012; Herb et al., 2014; Kulig, Joiner and Chang, 2015; Gribbin et al., 2016), while 13% (n=2/16) reported no significant differences between injured and uninjured groups (Drewes et al., 2009; Hein et al., 2012). Table 2.6 presents the breakdown of findings when specific injury types were group together.

Non-specific lower extremity injury types were reported in six studies and included running related injuries (n=5) (Ferber et al., 2005; Maclean et al., 2010; Meardon, Hamill and Derrick, 2011; Mann et al., 2015; Paquette, Milner and Melcher, 2016) and injury proneness (n=1) (James, Dufek and Bates, 2000). A significant between-group difference was reported in 33% (n=2/6) of these studies. When the injured group was compared to uninjured controls greater variability was evident in 33% (n=2/6) (James, Dufek and Bates, 2000; Maclean et al., 2010), reduced variability reported in 17% (n=1/6) (James, Dufek and Bates, 2000) and no difference in 67% (n=4/6) (Ferber et al., 2005; Meardon, Hamill and Derrick, 2011; Mann et al., 2015; Paquette, Milner and Melcher, 2016).

Table 2.6: Percentage of studies that showed greater, less or no variability when comparing injured and uninjured controls

| Study category (n) | Greater variability % (n) | Less variability % (n) | No difference in variability % (n) |
|---|---|---|---|
| **All Studies (22)** | 64(14) | 27(6) | 27(6) |
| **Single specific injury types (16)** | 75 (12) | 31 (5) | 13 (2) |
| CAI (5) (Drewes *et al.*, 2009; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Herb *et al.*, 2014; Hamacher, Hollander and Zech, 2016) | 60 (3) | 40 (2) | 20 (1) |
| ACLR (4) (van Uden *et al.*, 2003; Cordeiro *et al.*, 2015; Pollard *et al.*, 2015; Gribbin *et al.*, 2016) | 100(4) | 25 (1) | 0 |
| PFPS (2) (Heiderscheit, Hamill and Emmerik, 2002; Cunningham *et al.*, 2014) | 100 (2) | 0 | 0 |
| ITBS (2) (Miller *et al.*, 2008; Hein *et al.*, 2012) | 50 (1) | 50 (1) | 50 (1) |
| AGP (Edwards, Brooke and Cook, 2017) | 100 (1) | 0 | 0 |
| Hip OA (Chiu, Lu and Chou, 2010) | 100(1) | 0 | 0 |
| Patellar Tendinopathy (Kulig, Joiner and Chang, 2015) | 0 | 100 (1) | 0 |
| **Various lower limb injury** (6) (James, Dufek and Bates, 2000; Ferber *et al.*, 2005; Maclean *et al.*, 2010; Meardon, Hamill and Derrick, 2011; Mann *et al.*, 2015; Paquette, Milner and Melcher, 2016) | 33 (2) | 17 (1) | 67 (4) |
| **Injury by region** | | | |
| Hip (2) (Chiu, Lu and Chou, 2010; Edwards, Brooke and Cook, 2017) | 100 | 0 | 0 |
| Knee (10) (Heiderscheit, Hamill and Emmerik, 2002; van Uden *et al.*, 2003; Miller *et al.*, 2008; Maclean *et al.*, 2010; Hein *et al.*, 2012; Cunningham *et al.*, 2014; Cordeiro *et al.*, 2015; Kulig, Joiner and Chang, 2015; Pollard *et al.*, 2015; Gribbin *et al.*, 2016) | 80 (8) | 30 (3) | 10(1) |
| Ankle (5) (Drewes *et al.*, 2009; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Herb *et al.*, 2014; Hamacher, Hollander and Zech, 2016) | 60 (3) | 40 (2) | 20 (1) |
| **Pain** | | | |
| Pain (6) (Heiderscheit, Hamill and Emmerik, 2002; Chiu, Lu and Chou, 2010; Hein *et al.*, 2012; Kipp and Palmieri-Smith, 2012; Cunningham *et al.*, 2014; Cordeiro *et al.*, 2015) | 83 (5) | 0 | 17 (1) |
| Pain free (12) (James, Dufek and Bates, 2000; van Uden *et al.*, 2003; Ferber *et al.*, 2005; Miller *et al.*, 2008; Drewes *et al.*, 2009; Maclean *et al.*, 2010; Meardon, Hamill and Derrick, 2011; Brown, Bowser and Simpson, 2012; Kulig, Joiner and Chang, 2015; Mann *et al.*, 2015; Paquette, Milner and Melcher, 2016; Edwards, Brooke and Cook, 2017) | 42 (5) | 33 (4) | 42 (5) |
| **Analysis types** | | | |
| Continuous Measures (15) (Heiderscheit, Hamill and Emmerik, 2002; van Uden *et al.*, 2003; Ferber *et al.*, 2005; Miller *et al.*, 2008; Drewes *et al.*, 2009; Chiu, Lu and Chou, 2010; Maclean *et al.*, 2010; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Hein *et al.*, 2012; Cunningham *et al.*, 2014; Herb *et al.*, 2014; Pollard *et al.*, 2015; Gribbin *et al.*, 2016; Hamacher, Hollander and Zech, 2016) | 67 (10) | 20 (3) | 27 (4) |
| Continuous Measure Types | | | |
| Vector Coding (7) | 57 (4) | 29 (2) | 29 (2) |

Table 2.6: Percentage of studies that showed greater, less or no variability when comparing injured and uninjured controls (cont.)

| | | | |
|---|---|---|---|
| (Heiderscheit, Hamill and Emmerik, 2002; Ferber *et al.*, 2005; Maclean *et al.*, 2010; Cunningham *et al.*, 2014; Herb *et al.*, 2014; Pollard *et al.*, 2015; Gribbin *et al.*, 2016) | | | |
| Continuous Relative Phase (5) (van Uden *et al.*, 2003; Miller *et al.*, 2008; Drewes *et al.*, 2009; Chiu, Lu and Chou, 2010; Hein *et al.*, 2012) | 60 (3) | 0 | 40 (2) |
| Discrete Measures (10) (James, Dufek and Bates, 2000; Heiderscheit, Hamill and Emmerik, 2002; Miller *et al.*, 2008; Meardon, Hamill and Derrick, 2011; Kipp and Palmieri-Smith, 2012; Cordeiro *et al.*, 2015; Kulig, Joiner and Chang, 2015; Mann *et al.*, 2015; Paquette, Milner and Melcher, 2016; Edwards, Brooke and Cook, 2017) | 40 (4) | 30 (3) | 40 (4) |
| **Task** | | | |
| Cyclic (15) (Heiderscheit, Hamill and Emmerik, 2002; van Uden *et al.*, 2003; Ferber *et al.*, 2005; Miller *et al.*, 2008; Drewes *et al.*, 2009; Chiu, Lu and Chou, 2010; Maclean *et al.*, 2010; Meardon, Hamill and Derrick, 2011; Hein *et al.*, 2012; Cunningham *et al.*, 2014; Herb *et al.*, 2014; Mann *et al.*, 2015; Paquette, Milner and Melcher, 2016; Gribbin *et al.*, 2016; Hamacher, Hollander and Zech, 2016) | 53 (8) | 20 (3) | 40 (6) |
| Non-cyclic (7) (James, Dufek and Bates, 2000; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Cordeiro *et al.*, 2015; Kulig, Joiner and Chang, 2015; Pollard *et al.*, 2015; Edwards, Brooke and Cook, 2017) | 86 (6) | 43 (3) | 0 |
| **Variable** | | | |
| Kinematic (20) (Heiderscheit, Hamill and Emmerik, 2002; van Uden *et al.*, 2003; Ferber *et al.*, 2005; Miller *et al.*, 2008; Drewes *et al.*, 2009; Chiu, Lu and Chou, 2010; Maclean *et al.*, 2010; Brown, Bowser and Simpson, 2012; Hein *et al.*, 2012; Kipp and Palmieri-Smith, 2012; Herb *et al.*, 2014; Cunningham *et al.*, 2014; Kulig, Joiner and Chang, 2015; Mann *et al.*, 2015; Pollard *et al.*, 2015; Cordeiro *et al.*, 2015; Paquette, Milner and Melcher, 2016; Gribbin *et al.*, 2016; Hamacher, Hollander and Zech, 2016; Edwards, Brooke and Cook, 2017) | 60 (12) | 25 (5) | 30 (6) |
| Spatiotemporal (3) (Heiderscheit, Hamill and Emmerik, 2002; Meardon, Hamill and Derrick, 2011; Mann *et al.*, 2015) | 33 (1) | 0 | 67 (2) |
| Kinetic (3) (James, Dufek and Bates, 2000; Kipp and Palmieri-Smith, 2012; Edwards, Brooke and Cook, 2017) | 33 (1) | 33 (1) | 67 (2) |

Note: As some studies report significant findings of both lesser and greater variability or some studies utilized multiple methodological approaches (tasks, dependent variable types or analysis techniques) the total sum of percentages do not always equal 100% as one study may have been included for results under various subsections. CAI - chronic ankle instability, ACLR - anterior cruciate ligament reconstruction, PFPS - patellofemoral pain syndrome, ITBS - iliotibial band syndrome, AGP - athletic groin pain, OA -osteoarthritis.

**Findings by Region Location** Within the various subject groups, injury was spread across three lower limb regions (ankle, knee, hip). Injury at the knee joint was most commonly investigated. This was examined in 46% (n=10/22) of studies (Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Miller et al., 2008; Maclean et al., 2010; Hein et al., 2012; Cunningham et al., 2014; Cordeiro et al., 2015; Kulig, Joiner and Chang, 2015; Pollard et al., 2015; Gribbin et al., 2016) with significant findings evident in 90% of these (n=9/10). In the injured group greater variability was evident in 80% (n=8/10) of the studies (Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Miller et al., 2008; Maclean et al., 2010; Cunningham et al., 2014; Cordeiro et al., 2015; Pollard et al., 2015; Gribbin et al., 2016), reduced variability reported in 30% (n=3/10) (Miller et al., 2008; Kulig, Joiner and Chang, 2015; Gribbin et al., 2016) and no between-group differences were found in 10% (n=1/10) (Hein et al., 2012). The ankle was the next most commonly examined injury location with 23% (n=5/22) of all studies (Drewes et al., 2009; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Herb et al., 2014; Hamacher, Hollander and Zech, 2016). Significant between-group findings were evident in 80% of these (n=4/5) studies, with greater variability reported in the injured group in 60% (n=3/5) (Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Hamacher, Hollander and Zech, 2016), reduced variability in 40% (n=2/5) (Brown, Bowser and Simpson, 2012; Herb et al., 2014) and no difference in 20% (n=1/5) (Drewes et al., 2009). Injury at the hip region was examined in only 9% (n=2/22) of all studies with both of these (100%) (n =2/2) reporting that injured subjects demonstrated greater variability when compared to controls (Chiu, Lu and Chou, 2010; Edwards, Brooke and Cook, 2017). Groups consisting of injury across multiple lower limb regions (e.g. foot stress fracture, patellofemoral pain and hip pain) were examined in 23% (n=5/22) of studies and 80% reported no significant difference in variability between groups (Ferber et al., 2005; Meardon, Hamill and Derrick, 2011; Mann et al., 2015; Paquette, Milner

and Melcher, 2016). One study reported mixed findings with the injured group demonstrating increased and reduced variability when compared to uninjured controls (James, Dufek and Bates, 2000).

**Findings in Relation to Pain** Pain was reported in 27% (n=6/22) of the studies. Within these studies that reported pain, greater variability [83% (n=5/6)] was evident in the injured group compared to the uninjured group(Heiderscheit, Hamill and Emmerik, 2002; Chiu, Lu and Chou, 2010; Kipp and Palmieri-Smith, 2012; Cunningham et al., 2014; Cordeiro et al., 2015), while one study reported no between group differences (Hein et al., 2012). Two of these studies assessed subjects' pain levels using a visual analogue scale during the task performed (Heiderscheit, Hamill and Emmerik, 2002; Cunningham et al., 2014). Four studies reported pain, as assessed by orthopaedic examination (Hein et al., 2012) or a subjective questionnaire which included a subscale for pain (Chiu, Lu and Chou, 2010; Kipp and Palmieri-Smith, 2012; Cordeiro et al., 2015) but did not assess pain rating/level during the task.

Injured subject groups were reported as pain free at the time of testing in 55% (n=12/22) of studies (Mann et al. 2015; Meardon et al. 2011; Kulig et al. 2015; MacLean et al. 2010; van Uden et al. 2003; Miller et al. 2008; Brown et al. 2012; James et al. 2000; Ferber, Mcclay et al. 2005; Paquette et al. 2016; Edwards et al. 2016a; Drewes et al. 2009). Of these studies, significant findings were reported in 58% (n=7/12) of these studies. In the injured groups reporting no pain, greater movement variability in the injured group was evident in 42% (n=5/12) (James, Dufek and Bates, 2000; van Uden et al., 2003; Miller et al., 2008; Maclean et al., 2010; Brown, Bowser and Simpson, 2012), less variability in 33% (n=4/12) (James, Dufek and Bates, 2000; Miller et al., 2008; Brown, Bowser and Simpson, 2012; Kulig, Joiner and Chang, 2015) and no between-group difference in 42% (n=5/12)

(Ferber et al., 2005; Drewes et al., 2009; Meardon, Hamill and Derrick, 2011; Mann et al., 2015; Paquette, Milner and Melcher, 2016).

Four papers did not report on pain levels as part of the selection criteria or during the performance of the task (Herb et al., 2014; Pollard et al., 2015; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016).

**Findings by Analysis Type** In the reviewed studies, movement variability was examined in both continuous and discrete measures. Continuous measures refer to analysis of the entire biomechanical waveform (0-100%), while discrete measures refer to individual points on the waveform (e.g. peak knee moment; knee angle at peak ground reaction force; time to peak hip angle). It should be noted, that in both cases the analysis could be of a single joint/segment measure (e.g. knee moment) or between two joints/segments (e.g. hip-knee relative phase angle). Irrespective of whether a continuous or discrete measure was examined for variability, in line with the inclusion criteria of this review only studies employing linear statistical tools were utilized (e.g. standard deviation, coefficient of variation).

Continuous measures were examined for variability in 68% (n=15/22) of studies (Pollard et al. 2015; Gribbin et al. 2016; Kipp & Palmieri-Smith 2012; Heiderscheit et al. 2002; Cunningham et al. 2014; Chiu et al. 2010; MacLean et al. 2010; Brown et al. 2012; van Uden et al. 2003; Hamacher et al. 2016; Drewes et al. 2009; Ferber, Mcclay et al. 2005; Miller et al. 2008; Hein et al. 2012; Herb et al. 2014). Significant findings were reported in 73% (n=11/15). Greater variability in the injured group was evident in 67% (n=10/15) of the studies (van Uden et al., 2003; Miller et al., 2008; Chiu, Lu and Chou, 2010; Maclean et al., 2010; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Cunningham et al., 2014; Pollard et al., 2015; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016), reduced variability in 20% (n=3/15) (Brown, Bowser and Simpson, 2012;

Herb et al., 2014; Gribbin et al., 2016), and no difference observed between the two groups in 27% (n=4/15) (Heiderscheit, Hamill and Emmerik, 2002; Ferber et al., 2005; Drewes et al., 2009; Hein et al., 2012).

Variability was examined in continuous waveforms using various continuous measurement types including: vector coding (n=7) (Heiderscheit, Hamill and Emmerik, 2002; Ferber et al., 2005; Maclean et al., 2010; Cunningham et al., 2014; Herb et al., 2014; Pollard et al., 2015; Gribbin et al., 2016), continuous relative phase (n=5) (van Uden et al., 2003; Miller et al., 2008; Drewes et al., 2009; Chiu, Lu and Chou, 2010; Hein et al., 2012), ensemble curves of individual joint/segmental angles at each percent of the task cycle (n=2) (Brown, Bowser and Simpson, 2012; Hamacher, Hollander and Zech, 2016), and principal component analysis of discrete variables over pre-determined continuous time periods (e.g. 300ms) (n=1) (Kipp and Palmieri-Smith, 2012). Table 2.6 presents the breakdown of findings when variability was examined in continuous and discrete measures, and a further breakdown of the different types of continuous measures employed (e.g. vector coding and continuous relative phase).

The second approach used to quantify movement variability involved discrete point analysis, which examined maximum, minimum metrics to represent the whole waveform. This technique was used in 45% (n=10/22) of studies (James, Dufek and Bates, 2000; Heiderscheit, Hamill and Emmerik, 2002; Miller et al., 2008; Meardon, Hamill and Derrick, 2011; Kipp and Palmieri-Smith, 2012; Cordeiro et al., 2015; Kulig, Joiner and Chang, 2015; Mann et al., 2015; Paquette, Milner and Melcher, 2016; Edwards, Brooke and Cook, 2017) and significant findings were reported in 60% (n=6/10) of these studies. Greater variability was evident in 40% (n=4/10) (James, Dufek and Bates, 2000; Heiderscheit, Hamill and Emmerik, 2002; Cordeiro et al., 2015; Edwards, Brooke and Cook, 2017), reduced variability in 30% (n=3/10) (James, Dufek and Bates, 2000; Miller et al., 2008; Kulig, Joiner and Chang, 2015) and no difference reported in 40% (n=4/10) (Meardon, Hamill and

Derrick, 2011; Kipp and Palmieri-Smith, 2012; Mann et al., 2015; Paquette, Milner and Melcher, 2016) .

**Findings by Task** There were a wide variety of tasks used to examine movement variability including: running, walking, jumping, landing, side-step cutting and kicking a ball. To enable synthesis of findings the tasks were categorized into cyclic (end of one movement cycle is beginning of the next) or non-cyclic (distinct beginning and end) movements. Tasks that were cyclic in nature (running, walking or continuous single leg hopping) were utilized in 68% (n=15/22) of studies (Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Ferber et al., 2005; Miller et al., 2008; Drewes et al., 2009; Chiu, Lu and Chou, 2010; Maclean et al., 2010; Meardon, Hamill and Derrick, 2011; Hein et al., 2012; Cunningham et al., 2014; Herb et al., 2014; Mann et al., 2015; Paquette, Milner and Melcher, 2016; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016). Significant differences between the injured and uninjured groups were reported in 53% (n=9/15) of these studies. In the injured group greater variability was evident in 53% (n=8/15) (Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Miller et al., 2008; Chiu, Lu and Chou, 2010; Maclean et al., 2010; Cunningham et al., 2014; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016), reduced variability in 20% (n=3/15) (Miller et al., 2008; Herb et al., 2014; Gribbin et al., 2016) and no between-group difference reported in 40% (n=6/15) (Ferber et al., 2005; Drewes et al., 2009; Meardon, Hamill and Derrick, 2011; Hein et al., 2012; Mann et al., 2015; Paquette, Milner and Melcher, 2016). Movement variability was examined during a run condition in 87% (n=13/15) of these studies with 46% (n=6/13) reporting greater variability in the injured group when compared to controls (Heiderscheit, Hamill and Emmerik, 2002; Miller et al., 2008; Maclean et al., 2010; Cunningham et al., 2014; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016), while 8% (n=1/13) reported reduced variability (Miller et al., 2008), and 54% (n=7/13)

found no difference between groups (Ferber et al., 2005; Drewes et al., 2009; Meardon, Hamill and Derrick, 2011; Hein et al., 2012; Herb et al., 2014; Mann et al., 2015; Paquette, Milner and Melcher, 2016). Running was performed on a treadmill in 77% (n =10/13) of the studies (Heiderscheit, Hamill and Emmerik, 2002; Miller et al., 2008; Drewes et al., 2009; Maclean et al., 2010; Cunningham et al., 2014; Herb et al., 2014; Mann et al., 2015; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016; Paquette, Milner and Melcher, 2016), a runway in 15% (n =2/13) (Ferber et al., 2005; Hein et al., 2012) and a 300-meter indoor track in 8% (n =1/13) (Meardon, Hamill and Derrick, 2011). Run conditions were either set at a fixed (n=7) (Heiderscheit, Hamill and Emmerik, 2002; Drewes et al., 2009; Maclean et al., 2010; Hein et al., 2012; Herb et al., 2014; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016) or self-selected speed (n=7(Heiderscheit, Hamill and Emmerik, 2002; Ferber et al., 2005; Miller et al., 2008; Meardon, Hamill and Derrick, 2011; Cunningham et al., 2014; Mann et al., 2015; Paquette, Milner and Melcher, 2016). The average fixed speed was 10.4 km/h and the average self-selected speed for injured and uninjured control groups was 11.3 km/h. The length of the run condition ranged from 15 second trials to exhaustive 40 minute runs.

Non-cyclic tasks were utilized in 36% (n=7/22) of studies. A variety of different tasks were employed including a land (n=4/7) (James, Dufek and Bates, 2000; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Kulig, Joiner and Chang, 2015) side-step cut (n=2) (Pollard et al., 2015; Edwards, Brooke and Cook, 2017) and a soccer instep-kick (n=1) (Cordeiro et al., 2015). Significant differences in movement variability between injured groups and uninjured controls were identified in 100% (n =7/7) of these studies. Greater variability was evident in the injured group in 86% (n=6/7) (James, Dufek and Bates, 2000; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Cordeiro et al., 2015; Pollard et al., 2015; Edwards, Brooke and Cook, 2017) and reduced variability in

43% (n=3/7) (James, Dufek and Bates, 2000; Brown, Bowser and Simpson, 2012; Kulig, Joiner and Chang, 2015).

**Findings by Variables** Three types of dependent variable were utilized to examine movement variability between injured and uninjured subjects: kinematic, spatiotemporal and kinetic. Kinematic variables were examined in 91% (n=20/22) of studies (Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Ferber et al., 2005; Miller et al., 2008; Drewes et al., 2009; Chiu, Lu and Chou, 2010; Maclean et al., 2010; Brown, Bowser and Simpson, 2012; Hein et al., 2012; Kipp and Palmieri-Smith, 2012; Herb et al., 2014; Cunningham et al., 2014; Kulig, Joiner and Chang, 2015; Mann et al., 2015; Pollard et al., 2015; Cordeiro et al., 2015; Paquette, Milner and Melcher, 2016; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016; Edwards, Brooke and Cook, 2017) and 70% (14/20) of these reported significant findings. Greater variability was evident in the injured group in 60% (n=12/20) of these studies (van Uden et al., 2003; Miller et al., 2008; Chiu, Lu and Chou, 2010; Maclean et al., 2010; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Cunningham et al., 2014; Pollard et al., 2015; Cordeiro et al., 2015; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016; Edwards, Brooke and Cook, 2017), reduced variability in 25% (n=5/20) (Miller et al., 2008; Brown, Bowser and Simpson, 2012; Herb et al., 2014; Kulig, Joiner and Chang, 2015; Gribbin et al., 2016) and no between group differences was reported in 30% (n=6/20) (Heiderscheit, Hamill and Emmerik, 2002; Ferber et al., 2005; Drewes et al., 2009; Hein et al., 2012; Mann et al., 2015; Paquette, Milner and Melcher, 2016).

Spatiotemporal variables (e.g. foot contact time, stride time, stride length, stride frequency and flight time) were used to examine movement variability in 14% (n=3/22) of the studies. Greater variability in the injured group was reported in 33% of the studies (n=1/3) (Heiderscheit, Hamill and Emmerik, 2002), and no

between-group difference in 67% (n=2/3) (Meardon, Hamill and Derrick, 2011; Mann et al., 2015).

Kinetic variables (e.g. ground reaction forces, impulses and joint moments) were examined in 14% (n=3/22) of studies. One study report both significantly increased and decreased measures of variability in the injured group when compared to controls (James, Dufek and Bates, 2000), while no significant between-group difference was evident in 67% (n=2/3) (Kipp and Palmieri-Smith, 2012; Edwards, Brooke and Cook, 2017).

**Discussion**

To date, this is the first systematic review to investigate if there is a difference in movement variability between populations with a lower limb musculoskeletal injury compared with uninjured controls. The overall findings suggest that injured populations tend to deviate from 'normal' ranges of movement variability (as quantified in the uninjured control groups). 73% of studies reported significant between group differences in at least one measure of variability assessed, when comparing injured subject groups to uninjured controls (James, Dufek and Bates, 2000; Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Miller et al., 2008; Chiu, Lu and Chou, 2010; Maclean et al., 2010; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Cunningham et al., 2014; Herb et al., 2014; Kulig, Joiner and Chang, 2015; Pollard et al., 2015; Cordeiro et al., 2015; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016; Edwards, Brooke and Cook, 2017). The application of dynamic systems theory to help understand injury and movement variability has led to the hypothesis that an optimum range of variability exists for human movement, outside of which is associated with increased risk of injury (Stergiou, Harbourne and Cavanaugh, 2006; Hamill, Palmer and Van Emmerik, 2012). The focus in much of the literature has been that reduced variability has a greater association with injury (Hamill et al., 1999;

Davids et al., 2003), however, in direct contrast to this hypothesis the findings from this systematic review found a trend towards increased variability in injured groups, as 64% of studies reported significantly greater variability in the injured group when compared to the uninjured controls (James, Dufek and Bates, 2000; Heiderscheit, Hamill and Emmerik, 2002; van Uden et al., 2003; Miller et al., 2008; Chiu, Lu and Chou, 2010; Maclean et al., 2010; Brown, Bowser and Simpson, 2012; Kipp and Palmieri-Smith, 2012; Cunningham et al., 2014; Pollard et al., 2015; Cordeiro et al., 2015; Gribbin et al., 2016; Hamacher, Hollander and Zech, 2016; Edwards, Brooke and Cook, 2017). This was compared to 27% that reported less variability in the injured group. The varied findings from this review may be explained by the 'optimal' theory proposed by Hamill et al. (Hamill, Palmer and Van Emmerik, 2012), where either too little or too much movement variability may be related to increased risk of injury. This however should not be confused with the inverted U theory presented by Stergiou et al. (Stergiou, Harbourne and Cavanaugh, 2006), which examines the temporal structure of movement signals. The theory proposed by Hamill et al. (Hamill, Palmer and Van Emmerik, 2012) presents the relationship between motor performance and variability as a 'U' shaped association where either too little or too much is thought to have a detrimental impact on performance and may be associated with pathological populations. However, it is worth noting that to date no research has presented both significantly higher and lower variability in the injured group in comparison to the control group within the same variable.

There are a number of potential reasons why greater movement variability was identified in the injured groups. Firstly, as a causative risk factor for injury, it is possible that greater variability reflects a decrease in neuromuscular control leading to poorly controlled movement (Schmidt, 2013). This in turn may result in tissues (muscle, tendon, cartilage, bone) being subjected to unaccustomed strain and load, which if sustained overtime, or occurs at extremes in either force applied

or the associated motion, may lead to injury (Stergiou, Harbourne and Cavanaugh, 2006; Hamill, Palmer and Van Emmerik, 2012). If greater movement variability does represent a risk factor for injury then exercise interventions should focus on improving neuromuscular control. However, the retrospective design of the studies in the present review does not allow this causative relationship between altered movement variability and injury to be concluded.

Secondly, with injury pain provides an organismic influence, and the greater movement variability evident in injured populations may reflect an unstable compensatory movement mechanism being utilized in order to reduce loading on the sensitized and painful tissues (Tsao, Galea and Hodges, 2008; Madeleine and Madsen, 2009; Hodgesv and Tucker, 2011). Pain causes changes in the body at both central (spinal cord and brain) (Kotler et al., 1998; Hodgesv and Tucker, 2011) and peripheral levels (activation within and between muscles) (Hodgesv and Tucker, 2011) altering movement variability (Churchland, Afshar and Shenoy, 2006; Harbourne and Stergiou, 2009) and proprioceptive control of movement (Tsao, Galea and Hodges, 2008). This view is in line with dynamic systems theory, which suggests that movement patterns spontaneously arise through processes of self-organization as a result of several factors (e.g. task, environment, organismic) acting on the individual (Newell, Kugler and Emmerik, 1989; Davids et al., 2003). While such a change in movement mechanics may achieve a short-term goal of protection from further pain and or injury, this may not be ideal as a long-term movement solution; therefore interventions would have to aim at decreasing variability to 'normal' levels seen in uninjured populations (Hodgesv and Tucker, 2011).

A third explanation is that once the tissue damage resulting from an injury has healed and the pain has resolved, variability may represent the exploration of movement solutions by the neuromuscular system to re-optimize movement in the presence of altered neuromuscular capacity (e.g. reduced muscular strength)

and/or in the presence of pain induced changes to the body schema (the brain's representation of the body) (Schwoebel, 2001). In fact it has been previously observed that pain induced compensatory movement adaptions may even persist with recovery and the resolution of pain (Tsao, Galea and Hodges, 2008; Seay et al., 2011). If this latter explanation is the case, then rehabilitation interventions should not aim to decrease movement variability, as indicated above, as it potentially represents the natural recovery process. To determine if variability should be targeted in rehabilitation there is a clear need for intervention studies investigating the efficacy of increased/decreased variability-targeted rehabilitation. Furthermore, prospective research is required to conclusively determine the relationship between variability and injury.

Throughout this review, several studies produced contrasting findings for the same injury. For example greater (Hamacher, Hollander and Zech, 2016), less variability (Herb et al., 2014) and no difference in variability (Drewes et al., 2009) was identified at the ankle joint when examining chronic ankle instability. It is hard to identify exactly why these inconsistencies are evident, however it may be related to the different tasks examined (e.g. jump land, running, walking) and/or different measurement types utilised (e.g. continuous relative phase and vector coding). In fact even within individual studies when examining the same group for the same task, two authors identified both greater and less variability when examined at multiple joints/segments (Miller et al., 2008; Brown, Bowser and Simpson, 2012). This may reflect that in response to injury, the body reduces variability at certain joints to elicit a 'splinting effect' while increasing the variability at other joints to achieve the task outcome (Hodgesv and Tucker, 2011). These findings may indicate that the current application of dynamic systems theory to injuries requires revision.

**Subject Characteristics** Examining movement variability in a single specific injury group (e.g. chronic ankle instability) or injury at a specific region (e.g. hip, knee, ankle) appeared to be more sensitive in detecting an influence on variability than if examined in a non-specified group (e.g. general lower extremity running related injury comprising multiple injury types and/or across multiple injury regions). Where 75% of studies examining a single specific injury identified significant between group differences only 33% using a non-specific injury group reported significant findings. Similarly, 89% of the studies that examined one injury location found a significant difference in variability between groups compared to just 20% of the studies that examined a subject population consisting of several injury locations. This is perhaps not surprising as examining a heterogeneous group may mask findings between individuals within that group (Richter, O'Connor, et al., 2014b). For example, Ferber et al. (Ferber et al., 2005) included subjects with a variety of running related injuries, including patellofemoral pain syndrome, and reported no significant between group difference; while Heiderscheit et al. (Heiderscheit, Hamill and Emmerik, 2002) and Cunningham et al. (Cunningham et al., 2014) included only subjects with patellofemoral pain syndrome and both studies reported significant between-group differences.

**Analysis Type** Findings from this review possibly favour examining variability in continuous measures, with 73% of studies examining a continuous waveform reporting significant differences between injured subjects and uninjured controls. This is in comparison to 60% of the studies examining discrete point data. In accordance with this finding, Kipp & Palmieri-Smith (Kipp and Palmieri-Smith, 2012) examined variability in both discrete and continuous measures and found that differences in variability between injured and uninjured groups was only detectable in continuous measures. Variability was examined in a number of continuous data analysis techniques within this review; these

included the variation in continuous relative phase, vector coding and principle component analysis. With the variety of approaches utilized no trend was identified regarding their sensitivity. Further, no study directly compared different continuous analysis techniques.

**Task** When tasks were categorized into cyclic or non-cyclic movements, the non-cyclic tasks appeared more sensitive at detecting differences in variability between injured and uninjured groups. All studies examining non-cyclic tasks reported significant between group differences. In comparison, only half of those that examined cyclic tasks reported significant between group differences. There are possibly three explanations for this finding. Firstly, this may simply represent a methodological/statistical phenomenon, in which the methods utilized are more suitable at detecting differences during non-cyclic tasks in comparison to cyclic tasks. Secondly, the greater detection of variability in non-cyclic tasks may be related, at least in part, to cyclic tasks, such as walking or running, being practiced more often (Stergiou and Decker, 2011). As such, at the time of testing subjects may have already explored alternative movement strategies for these cyclic tasks and have adopted a less variable movement solution. In contrast, non-cyclic tasks such as landings and cutting maneuverers are generally less practiced. This may result in alternative movement strategies still being explored at the time of testing resulting in greater variability (Bartlett, 2008). Finally, non-cyclic tasks (e.g. landing, change of direction) typically demonstrate greater ground reaction forces and loading than cyclic tasks (e.g. running, hopping). The greater detection of variability in non-cyclic tasks may therefore be reflective of greater compensation being utilized in comparison to cyclic tasks in order to offload the injured tissues.

It is worth noting, but probably unrelated to the previous point, that the median number of strides examined for cyclic tasks was 14 and the median number of trials for non-cyclic tasks was 5 with an interquartile range of 25.50

and 6.25 respectively. Within this review, no study provided justification for the number of trials examined. Previous research examined the number of trials required to reliably examine movement variability during walking (Sangeux et al., 2016; Hafer and Boyer, 2017) and running (Hafer and Boyer, 2017) identifying that up to 20 trials are required for walking and 8 trials are required for running, similar to the median number of trials examined by studies in this review. However, caution should be applied here in adopting the findings of Sangeux et al. (Sangeux et al., 2016) and Hafer and Boyer (Hafer and Boyer, 2017) as the number of trials required has been shown to be task dependent (Hafer and Boyer, 2017), and is likely specific to the population being examined, and these studies examined only healthy participants. Future research should therefore look to investigate the number of trials needed to reliably examine movement variability during specific tasks and in specific populations. This should ideally be conducted not with just statistical reliability, as in Sangeux et al. (Sangeux et al., 2016) and Hafer and Boyer (Hafer and Boyer, 2017) whereby the number of trials was determined based on the magnitude of variability staying within a given range (e.g. within 10% of a 15-stride mean or 10% relative precision) as the number of trials is increased. Rather the number of trials required should be determined with regard to how many trials are required to detect differences with an appropriately high effect-size in the amount of variability between subjects of interest (e.g. in injured subjects with chronic ankle instability vs. uninjured controls).

When cyclic tasks were examined, studies which utilised treadmill tasks appeared to be more sensitive at detecting differences in variability between injured and uninjured groups with 70% of the studies reporting significant between group differences. In comparison, none of the studies which examined over-ground cyclic tasks reported significant between group differences. This is possibly related to greater variations related to the over-ground task itself masking between group differences in variability.

**Variables** Within the reviewed studies, kinematic, kinetic and spatiotemporal measures were all utilized as dependent measures to compare movement variability between injured subjects and uninjured controls. Kinematic measures were the most commonly examined in 91% of studies and 70% of these reported a significant between-group difference. In comparison, spatiotemporal and kinetic measures were each only examined in 14% of studies and 33% of these studies identified a significant difference between the two groups. When studies compared both kinetic and kinematic variables significant between-group differences were only evident in kinematic variables (Kipp and Palmieri-Smith, 2012; Edwards, Brooke and Cook, 2017). These findings at present would suggest that kinematic variability might be the most sensitive to differences between groups. This was somewhat surprising as kinetic measures reflect more closely the forces associated with injuries. This is difficult to explain as there is generally greater intra-subject variability in kinetic measures (Winter, 2009). Previous research has demonstrated that kinetic measures also have proportionally higher between-subject variability (Winter, 2009) which inhibits the ability to detect difference between groups. However, to date no research has examined the variability in kinetic variability measurements so it remains uncertain why kinematic variability might be the most sensitive to differences between groups.

These findings have numerous methodological implications. Firstly, when examining movement variability, it may be more appropriate for researchers to investigate a single specific injury type thus avoiding any possible masking of findings. Secondly, continuous analysis of data may be more sensitive than discrete-point analysis in detecting significant between-group differences. Finally, kinematic movement variability measured during non-cyclic tasks appeared to be most consistent in detecting significant between-group differences.

**Limitations of the Review** Given the large heterogeneity in study methodologies and metrics examined, this review was limited to a qualitative analysis of the literature. The qualitative approach adopted here results in papers being categorized as either finding greater variability or less variability if any one measure, of all the measures assessed, has greater or less variability, respectively. In contrast, for a paper to be categorized as having no effect on variability, none of the measures assessed must differ between the injured and uninjured controls. This introduces a bias in the reporting process that needs to be considered when interpreting the findings. For full report of findings please see appendix A.

Another limitation to this review is the underlying assumption that a 'normal' magnitude of variability exists and is associated with healthy individuals. As with all cross comparison studies it may be that the 'healthy' subjects also have an abnormal level of variability, thereby predisposing them to injury, but have not been exposed to the volume or intensity required to develop an injury.

Secondary to the above point it is not possible to state an overall 'normal' range of variability within this qualitative review because of the wide variety of tasks, injury groups and analysis methods employed.

**Limitations of the studies reviewed** Limitations of the studies reviewed include; poor control and/or reporting of possible confounding factors (including fatigue, gender, pain, running speed), biased examination of multiple comparisons without controlling for family-wise errors and a lack of justification for the number of trials analysed when examining between-trial variability. The later point is of importance as it is currently unknown how many trials are required to obtain reliable and meaningful measures. Other limitations include the use of the coefficient of variation metric in several studies (Heiderscheit, Hamill and Emmerik, 2002; Meardon, Hamill and Derrick, 2011; Brown, Bowser and Simpson, 2012; Kulig, Joiner and Chang, 2015; Mann et al., 2015) which may result in

findings of variability that are artificially inflated when the variable examined are of a small magnitude. Also the plethora of measurement types utilized within this review impedes the cross comparison and synthesis of findings between studies. A consensus should be reached for the standardization of analysis types. Finally, the lack of large normative databases currently limits the utility of variability as a clinical measure. To encourage the adoption of best research practice, and allow the use of variability as a clinical measure, open access databases should be encouraged. This would not only enhance the transparency of research but also allow the investigation and direct comparison of these different methodologies.

**Summary of systematic review**

The overall findings from this review suggest that deviation from normal ranges of movement variability may be associated with injury. Interestingly, a trend was identified with injured populations exhibiting greater movement variability when compared to uninjured controls. This trend was evident in the injured populations group reporting pain and ligamentous injury and as such future research should explore the association between pain and variability and sensorimotor control and variability. There is a clear need to repeat many of the studies within this review, using appropriate methodologies, to determine if there is a consistency in the findings of variability between those with and without injury. In order to enhance the sensitivity to detect between group differences in variability future research should consider examining variability using a subject group with a clearly defined injury type and/or injury location. Furthermore, it would appear advantageous to examine non-cyclic tasks utilizing continuous waveform methods of analysis, with a focus on kinematic measures. Finally, prospective research needs to be conducted to determine if alterations in movement variability precedes or follows an injury.

**Implication of the systematic review for this thesis** Athletic groin pain (AGP) is a common injury, typically associated with sports involving repetitive agility tasks (Werner et al., 2009; Orchard, Seward and Orchard, 2013; Thorborg et al., 2017). Given the association between repetitive loading and chronic overuse injuries such as AGP, there has been a growing interest in the functional role movement variability may have with respect to injuries such as AGP (Stergiou and Decker, 2011; Hamill, Palmer and Van Emmerik, 2012; Baida et al., 2017). The findings from this review suggest that movement variability is affected by injury and as such it's investigation in AGP is warranted.

In line with the findings from this review, variability in AGP will be examined using a non-cyclic task, utilizing continuous waveform methods of analysis with a focus on kinematic measures. Furthermore, as per the recommendations of this review, this thesis will investigate the number of trials required to detect differences with an appropriately high effect-size in the amount of variability between those with and without AGP during a task of interest.

## 2.6 Exercise Intervention

Exercise interventions have been shown to be an effective means of rehabilitating AGP as concluded by several systematic reviews (Jansen et al., 2008; Machotka, Kumar and Perraton, 2009; Almeida et al., 2013; Charlton et al., 2017). Indeed, when comparing surgical and exercise interventions, there are no clear differences between these two approaches in terms of return to play time or rate (King et al., 2015).

Exercises involving external load, particularly targeted at the hip and abdominal musculature, are commonly prescribed for the treatment of AGP (Charlton et al., 2017). These exercise interventions have been coupled with rest and active recovery (Holmich et al., 1999) and with manual therapy (Weir et al., 2011) with

good long term outcomes (Holmich, Nyvold and Larsen, 2011). However, as noted by King et al., (2018) the focus of these studies on patients presenting with a single anatomical presentation (Holmich et al., 1999; Weir et al., 2011) may reduce their generalisability in treating athletes presenting with other entities. These exercise rehabilitation programs may also be limited by their time based (rather than criterion based) approach to rehabilitation and a focus on localised low load exercise (Holmich et al., 1999; Weir et al., 2011).

Franklyn-Miller et al., (2016) contend that AGP is caused by an overload of the anterior pubic area (pubic symphysis and surrounding tissues), with various structures becoming painful in direct response to this loading or in an attempt to stabilize the region during dynamic loading tasks. Therefore, a more appropriate and specific means of rehabilitation for AGP would include an individualised exercise programme focused on intersegmental control during plyometric and multi directional tasks. This form of rehabilitation was presented by King et al., (2018) who reported a 73% RTP rate in 9.9 $\pm$3.4 weeks in a cohort of 205 patients. This compare favourably with anatomically specific rehabilitation protocols used by Holmich et al., (1999) [68% RTP, 18.5 weeks] and Weir et al., (2011) [48% RTP, 17.3 weeks]. While it should be noted that both of the aforementioned studies are randomised control trials (Holmich et al., 1999; Weir et al., 2011), and therefore are not directly comparable to the study by King et al., (2018), it is plausible that quicker return to play time reported in King et al., (2018) was due to the rehabilitation being more targeted in nature with a focus on whole body movements. The following section will outline the rehabilitation program presented by King et al., (2018) and utilised within this thesis. For a more thorough justification and detailed description of this intervention please see King et al., (2018) where the intervention has been published in accordance with the TIDieR (template for intervention description and replication) checklist

and guide (Hoffmann et al., 2014). In addition, a summary of the exercise protocol is included in Appendix C.

**Rehabilitation program**

Within this thesis, where an exercise intervention was utilised, all subjects with AGP undertook a rehabilitation programme focused on control of the hip, pelvis and trunk during dynamic loading tasks. The intervention involved three levels of progression ( Figure 2.6).

Level 1 addressed inter-segmental control and strength, level 2 involved linear running mechanics and increasing linear running load tolerance and level 3 addressed multidirectional mechanics and the transition back to high intensity sprinting. This program was completed 4 days per week alternating between strength and running drills. Advancement through these levels was based on the examining physiotherapist's subjective assessment of the subjects' control, absence of pain during tasks of the preceding level alongside the competence-based assessment of activity. The programme was unsupervised but a physiotherapist assessed each patient's progress at regular intervals. This approach ensured the speed of programme progression was individualised as the individual competency and symptom levels allowed, ensuring the most appropriate exercise selection and recovery time (King et al., 2018). Patients who demonstrated symptom free completion of the Linear B running programme and multidirectional drills at maximum intensity were deemed sufficiently rehabilitated to be cleared to return to play.

**Summary**

Exercise interventions are an effective means of AGP rehabilitation. Rather than focusing on individual anatomical locations with localised low load exercise, it is possible that a rehabilitation programme focused on control of the hip, pelvis

Figure 2.6: Components of AGP rehabilitation and key performance indicators for progression (King et al. 2018)

and trunk during dynamic loading tasks may be a more appropriate and specific means of AGP rehabilitation. The above section provides an outline of how such a rehabilitation program was structured within this thesis.

## 2.7 Biomechanical screening tests

As suggested by van Mechelen, Hlobil, & Kemper, (1992), following establishment of the epidemiology surrounding an injury, it is important to investigate the risk factors associated with it. Given that injuries are directly caused by excessive relative loading, a 3D biomechanical assessment is an essential step for the understanding of injury risk factors and the subsequent development of injury prevention programmes (Bahr and Krosshaug 2005, Finch 2006). Various test actions have been used to either screen for predisposition to injury or exam-

ine the effectiveness of rehabilitation. Test actions should ideally replicate the biomechanical demands of the actions associated with the injury (Marshall et al. 2016). The following section will review the mechanics of the two test actions utilised within this thesis, namely the lateral hurdle hop and the running cut completed by uninjured cohorts. While this information is important both to understand the suitability of various test actions and provide normative data when exploring AGP mechanics, care is required when interpreting the data in this chapter. Many of the studies utilising conventional gait models (Davis et al., 1991) may exhibit the presence of cross talk. Cross talk occurs when markers are misplaced and results in modifying the orientation of the local coordinate systems. This is particularly evident at the knee where the angles achieved in the frontal plane often exceed the 7° range of motion that is physiologically possible (Baudet et al., 2014).

The running cut is a commonly examined task in biomechanics due to its high multi-planar loading (Pollard, Sigward and Powers, 2007; Beaulieu, Lamontagne and Xu, 2008; Sanna and O'Connor, 2008; Havens and Sigward, 2015; Edwards, Brooke and Cook, 2017) and ecological validity to injuries that tend to occur during field sports. The hurdle hop will also be reviewed, as like the running cut, it involves much higher frontal plane loading than other commonly utilised screening tasks [e.g. the vertical drop jump (Maloney et al. 2015)]. Finally, given the small number of studies which have biomechanically examined lateral hurdle hopping, the single leg drop landing will also be reviewed given its similarities to the eccentric phase of the lateral hurdle hop landing. The purpose of this section is to explore the test actions utilised within this thesis and present normative data. This section is not intended to be an extensive review of the published literature, rather it is a presentation of randomly selected research to provide a good representation of published normative data.

### 2.7.1 The running cut

As noted, it is important that biomechanical screening replicate the demands of the actions associated with the injury (Marshall et al. 2016). Athletic groin pain is common in sports involving rapid agility tasks (Thorborg et al. 2017, Orchard, Seward and Orchard 2013, Werner et al. 2009). As such running cut tasks may be an ecologically valid means of assessing the mechanics of AGP in comparison to uninjured controls. The following section will outline the mechanics of the running cut in the sagittal plane before reviewing the biomechanics of the running cut task as presented in the literature. Within the summary text when more than one study has presented the same variable, results will be presented as ranges. In contrast when only study has presented results for a specific variable, the result presented will be the mean and standard deviation.

#### Outline of the Running Cut

The participants run toward a marker placed on the floor, make a single complete foot contact on the force plate and perform a cutting action to change direction. Within this current thesis, the cutting angle is 110° with respect to the approach direction. Participants are instructed to run and complete the total cutting task as quickly as possible (Figure 2.7).

Typical sagittal plane biomechanics for the pivot ground contact of the running cut is presented in Figure 2.8 below. The hip extends through a small range of motion for the duration of the task. The knee flexes from approximately 10% to 50% of the ground contact. The knee then extends until around 90% of the ground contact before it flexes slightly before toe off. Similarly, the ankle flexes from approximately 10% to 70% of the ground contact. At this point the ankle joint rapidly extends until toe off.

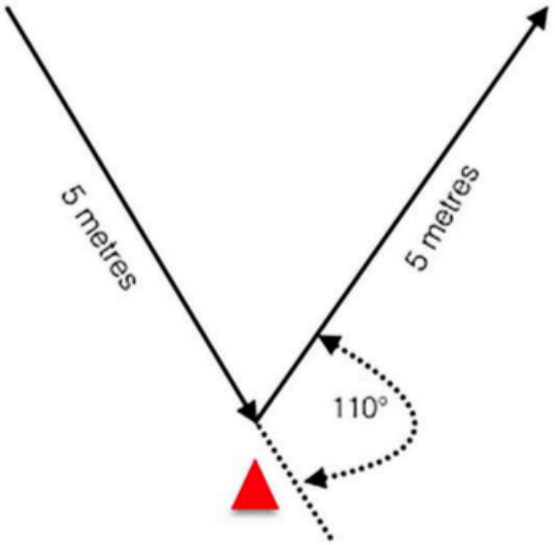


Figure 2.7: Illustration of the 110 ° cut task

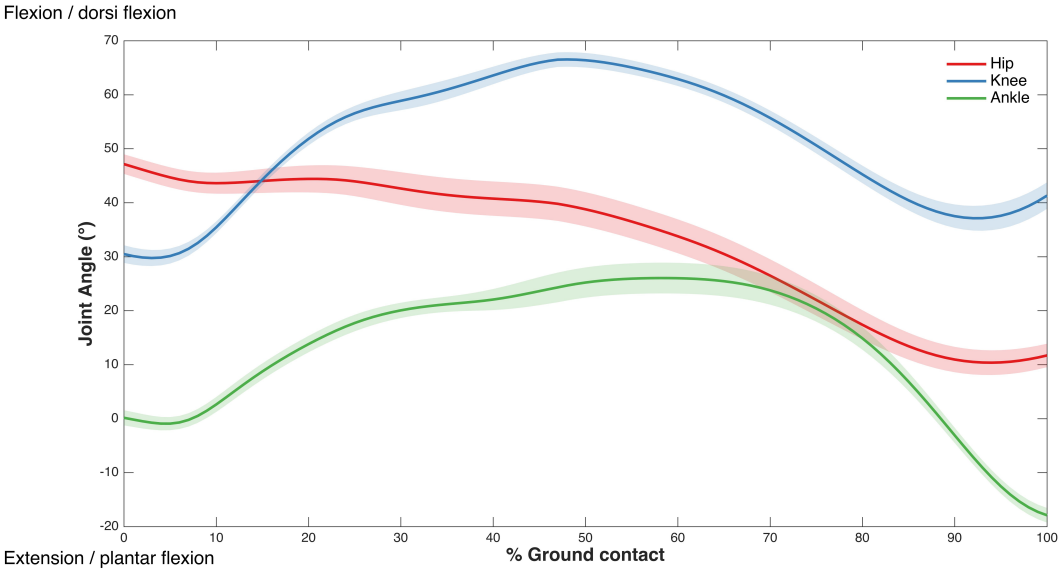


Figure 2.8: Sagittal plane kinematics of the hip, knee and ankle during a running cut. Shaded band indicates standard error.

Of the nine studies reviewed, five studies investigated a 45° cutting task (Havens and Sigward 2015, Edwards, Brooke and Cook 2017, Pollard, Sigward and Powers 2007, Beaulieu, Lamontagne and Xu 2008, Sanna and O'Connor 2008). One study examined each of 30° (Kristianslund et al. 2014), 35 - 40° (McLean, Huang and Van Den Bogert 2005), 40° (Sigward and Powers 2006), 90° (Havens and Sigward 2015) and 110° (Marshall et al. 2015) cutting task. The majority of studies examined an anticipated cutting task with only three studies examining an unanticipated task (Edwards, Brooke and Cook 2017, Pollard, Sigward and Powers 2007, Beaulieu, Lamontagne and Xu 2008)

Kinematic results for the running cut are presented in Tables 2.7 to 2.10 below. At the thorax at initial contact, there is between 6.1° and 15.18° of contralateral side flexion and 7.1 ± 14.5° of ipsilateral rotation. At the hip, there is between 32.1 and 50.6° of flexion, between 3.0° and 23.94 ° of abduction and 3.9° to 12.96° of internal rotation. At the knee there is between 14.67 ° and 28.1° of flexion on initial contact, 0.9° to 8.3° of abduction and 0.8° and 0.17° of external rotation. At the ankle at initial contact there is between 0.8° and 12.86° of plantar-flexion, 9.4 ± 5.3° of ankle inversion and 8.7 ± 6.8° of ankle adduction. During the whole body eccentric phase, the trunk is anteriorly flexed 30.5 ± 5.8°, ipsilateral side flexed -21.0 ± 7.9 ° and internally rotated 11.8 ± 6.6°. The pelvis is flexed 2.2 ± 5.1°, contralateral dropped -15.0 ± 5.9° and externally rotated -11.1 ± 13.1°. The hip flexes to 45.1 ± 11.9°, abducts to 17.9 ± 6.7° and internally rotates to 22.4 ± 10.1°. At the knee there is 57.4 ± 6.0° of flexion, -7.5 ± 5.0° of adduction and 21.2 ± 9.4° of internal rotation. At the ankle there is 11.1 ± 7.6° of plantar-flexion, 5.4 ± 2.4° of ankle eversion and 33.5 ± 13.2 of external rotation. During the total ground contact, the hip flexes to 54.1 ± 11.0°, abducts to between 9.07 and 33.1° and internally rotates between 3.58 and 14.6°. At the knee there is between 57.36 and 63.1° of flexion, between 1.53 and 12.1° of knee abduction and between 6.07

Table 2.7: Angles (°) at initial contact during the running cut

| Author | Joint | Unanti (cm) | Cut Angle (°) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Kristianslund et al. 2014 | **Trunk** | N | 30° | Female | | 6.1 ± 8.6 | 7.1 ± 14.5 |
| Havens and Sigward 2015 | **Trunk** | N | 45° | Mixed | | 9.25 ± 4.78 | |
| Havens and Sigward 2015 | **Trunk** | N | 90° | Mixed | | 15.18 ± 6.71 | |
| | | | | | | | |
| Havens and Sigward 2015 | **Hip** | N | 45° | Mixed | 46.06 ± 6.82 | 3.90 ± 7.07 | 12.96 ± 6.23 |
| Havens and Sigward 2015 | **Hip** | N | 90° | Mixed | 32.08 ± 11.05 | 23.94 ± 4.85 | 10.14 ± 7.88 |
| Kristianslund et al. 2014 | **Hip** | N | 30° | Female | | 17.2 ± 5.9 | 3.9 ± 8.3 |
| Sanna and O'Connor 2008 | **Hip** | N | 45° | Female | 50.6 ± 9.8 | 3.0 ± 6.0 | 8.8 ± 6.5 |
| | | | | | | | |
| Havens and Sigward 2015 | **Knee** | N | 45° | Mixed | 25.21 ± 6.97 | | |
| Havens and Sigward 2015 | **Knee** | N | 90° | Mixed | 14.67 ± 6.35 | | |
| Beaulieu et al., 2008 | **Knee** | Y | 45° | Male | 15.60 ± 6.11 | 1.28 ± 6.22 | -0.17 ± 9.27 |
| Kristianslund et al. 2014 | **Knee** | N | 30° | Female | 24.0 ± 6.1 | 8.3 ± 4.4 | |
| Sanna and O'Connor 2008 | **Knee** | N | 45° | Female | 28.1 ± 4.2 | 0.9 ±1.4 | -0.8 ± 4.9 |
| | | | | | | | |
| Havens and Sigward 2015 | **Ankle** | N | 45° | Mixed | 4.48 ± 8.86 | | |
| Havens and Sigward 2015 | **Ankle** | N | 90° | Mixed | 12.86 ± 11.85 | | |
| Sanna and O'Connor 2008 | **Ankle** | N | 45° | Female | 0.8 ± 9.4 | 9.4 ± 5.3 | 8.7 ± 6.8 |

Unanti = Unanticipated, N/Y = no/yes, Flexion/Dorsiflexion = +ive, Abduction/inversion/contra lateral tilt = +ive, Internal Rotation =+ive

and 22.91° of internal rotation. At the ankle there is $1.5 \pm 4.9°$ of ankle eversion. Range of motion during the cutting task is presented by one study in Table 2.13

Table 2.8: Peak Angles (°) during the eccentric phase for the running cut

| Author | Joint | Unanti (cm) | Cut Angle (°) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Marshall et al., 2015 | **Trunk** | N | 110° | Male | 30.5 ± 5.8 | -21.0 ± 7.9 | 11.8 ± 6.6 |
| Marshall et al., 2015 | **Pelvis** | N | 110° | Male | 2.2 ± 5.1 | -15.0 ± 5.9 | -11.1 ± 13.1 |
| Marshall et al., 2015 | **Hip** | N | 110° | Male | 45.1 ± 11.9 | 17.9 ± 6.7 | 22.4 ± 10.1 |
| Marshall et al., 2015 | **Knee** | N | 110° | Male | 57.4 ± 6.0 | −7.5 ± 5.0 | 21.2 ± 9.4 |
| Marshall et al., 2015 | **Ankle** | N | 110° | Male | 1.9 ± 0.4 | -0.7 ± 0.2 | 0.1 ± 0.1 |

Unanti = Unanticipated, N/Y = no/yes, Flexion/Dorsiflexion = +ive, Abduction/inversion/contra lateral tilt = +ive, Internal Rotation =+ive

Table 2.9: Peak angles (°) during the running cut

| Author | Joint | Unanti (cm) | Cut Angle (°) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| McLean et al., 2005 | **Hip** | N | 35 -40° | Male | 54.1 ± 11.0 | 33.1 ± 8.9 | 14.6 ± 7.8 |
| Pollard et al., 2007 | **Hip** | Y | 45° | Male | | 9.07 ± 7.2 | 3.58 ± 8.9 |
| McLean et al., 2005 | **Knee** | N | 35 -40° | Male | 63.1 ± 9.5 | 12.1 ± 4.5 | 19.2 ± 5.9 |
| Pollard et al., 2007 | **Knee** | Y | 45° | Male | | 1.53 ± 6.0 | 6.07 ± 5.9 |
| Beaulieu et al., 2008 | **Knee** | Y | 45° | Male | 57.36 ± 5.01 | 5.26 ± 11.28 | 22.91 ± 6.92 |
| McLean et al., 2005 | **Ankle** | N | 35 -40° | Male | | -1.5 ± 4.9 | |

Unanti = Unanticipated, N/Y = no/yes, Flexion/Dorsiflexion = +ive, Abduction/inversion/contra lateral tilt = +ive, Internal Rotation =+ive

Table 2.10: Angle range of motion (°) for the running cut

| Author | Joint | Unanti (cm) | Cut Angle (°) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Sanna and O'Connor 2008 | **Hip** | N | 45° | Female | 59.2 ± 4.7 | 9.9 ± 3.8 | 12.6 ± 3.6 |
| Sanna and O'Connor 2008 | **Knee** | N | 45° | Female | 25.1 ± 7.3 | 4.5 ± 2.3 | 13.7 ± 2.4 |
| Sanna and O'Connor 2008 | **Ankle** | N | 45° | Female | 25.7 ± 10.8 | 15.5 ± 6.5 | 9.5 ± 5.2 |

Unanti = Unanticipated, N/Y = no/yes, Flexion/Dorsiflexion = +ive, Abduction/inversion/contra lateral tilt = +ive, Internal Rotation =+ive

Moments are presented in in Tables 2.11 to 2.13. Four studies presented peak moments during the total ground contact of running cut. All but one study (Sanna and O'Connor 2008) presented their findings as normalised to body mass. For comparative purposes, the findings from Sanna and O'Connor (2008) have been normalised by the mean mass of the study participants. At the hip there was between 3.11 and 4.65 Nm/Kg of extensor moments, between -0.96 and 3.20 Nm/Kg of hip adductor moments and between -0.47 and 1.64 Nm/Kg internal rotation moments. At the knee there was between 1.4 and 3.06 Nm/Kg of extensor moments, between 0.006 and 1.03 Nm/Kg of knee adductor moments and -0.09 and 0.18 Nm/Kg internal rotation moments. At the ankle there was between 2.35 and 3.04 Nm/Kg of plantar-flexor moments, $0.81 \pm 0.22$ Nm/Kg of ankle invertor moments and $0.95 \pm 0.22$ Nm/Kg of external rotation moments. During the eccentric/ weight acceptance phase of the ground contact, at the hip there was between 4.0 and 4.67 Nm/Kg of extensor moments, between 1.28 and 3.6 Nm/Kg of hip adductor moments and -1.3 and 1.99 Nm/Kg internal rotation moments. At the knee there was between 2.6 and 3.51 Nm/Kg of extensor moments, between 0.94 and 2.5 Nm/Kg of knee adductor moments and 0.4 and 0.70 Nm/Kg internal rotation moments. At the ankle there was between 1.67 and 1.9 Nm/Kg of plantar-flexor moments, between -0.7 and 0.39 Nm/Kg of ankle evertor moments and 0.1 and 0.41 Nm/Kg internal rotation moments.

Table 2.11: Peak Moments (Nm/Kg) during the running cut

| Author | Joint | Unanti (cm) | Cut Angle (°) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Pollard et al., 2007 | **Hip** | Y | 45° | Male | | -0.96 ± 0.3 | -0.47 ± 0.4 |
| Havens and Sigward 2015 | **Hip** | N | 45° | Mixed | 4.65 ± 1.41 | 1.40 ±1.46 | 1.64 ± 0.90 |
| Havens and Sigward 2015 | **Hip** | N | 90° | Mixed | 3.11 ± 1.10 | 3.20 ± 1.27 | 1.60 ± 0.56 |
| Sanna and O'Connor 2008 | **Hip** | N | 45° | Female | 4.09 ± 0.88† | 1.59 ± 0.77† | 0.42 ± 0.18† |
| Pollard et al., 2007 | **Knee** | Y | 45° | Male | | 0.31 ± 0.1 | -0.09 ± 0.1 |
| Sigward and Powers 2006 | **Knee** | N | 40° | Male | 2.1 ± 0.8 | 0.006 ± 0.3 | |
| Sigward and Powers 2006 | **Knee** | N | 40° | Female | 1.4 ± 0.7 | 0.43 ± 0.5 | |
| McLean et al., 2005 | **Knee** | N | 35 -40° | Male | | 0.42 ± 0.11 | |
| Havens and Sigward 2015 | **Knee** | N | 45° | Mixed | 2.67 ± 0.73 | | |
| Havens and Sigward 2015 | **Knee** | N | 90° | Mixed | 3.06 ± 0.60 | | |
| Sanna and O'Connor 2008 | **Knee** | N | 45° | Female | 2.31 ± 0.51† | 1.03 ± 0.40† | 0.18 ± 0.02† |
| Havens and Sigward 2015 | **Ankle** | N | 45° | Mixed | 3.04 ± 0.53 | | |
| Havens and Sigward 2015 | **Ankle** | N | 90° | Mixed | 2.58 ± 0.58 | | |
| Sanna and O'Connor 2008 | **Ankle** | N | 45° | Female | 2.35 ± 0.28† | -0.81 ± 0.22† | -0.95 ± 0.22† |

Moments reported as internal moments. Unanti = Unanticipated, N/Y = no/yes '†' = value normalised by mean weight (76.2 kg), Extension/Plantarflexor = +ive, Adductor/eversion = +ive, Internal Rotator =+ive

Table 2.12: Moments (Nm/Kg) during the eccentric phase / weight acceptance for the running cut

| Author | Joint | Unanti (cm) | Cut Angle (°) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Marshall et al., 2015 | **Hip** | N | 110° | Male | 4.0 ± 1.4 | 3.6 ± 1.4 | -1.3 ± 0.5 |
| Edwards et al., 2017 | **Hip** | Y | 45° | Male | 4.67 ± 1.78 | 1.28 ± 0.52 | 1.99 ± 0.73 |
| | | | | | | | |
| Marshall et al., 2015 | **Knee** | N | 110° | Male | 2.6 ± 0.5 | 2.5 ± 1.0 | 0.4 ± 0.1 |
| Edwards et al., 2017 | **Knee** | Y | 45° | Male | 3.51 ± 0.52 | 0.94 ± 0.30 | 0.70 ± 0.27 |
| | | | | | | | |
| Marshall et al., 2015 | **Ankle** | N | 110° | Male | 1.9 ± 0.4 | -0.7 ± 0.2 | 0.1 ± 0.1 |
| Edwards et al., 2017 | **Ankle** | Y | 45° | Male | 1.67 ± 0.93 | 0.39 ± 0.26 | 0.41 ± 0.21 |

Moments reported as internal moments. Unanti = Unanticipated, N/Y = no/yes, Extension/Plantarflexor = +ive, Adductor/eversion = +ive, Internal Rotator =+ive

Table 2.13: Mean joint power (W/kg) for the running cut

| Author | Joint | Unanti (cm) | Cut Angle (°) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Havens and Sigward 2015 | **Hip** | N | 45° | Mixed | 6.66 ± 3.50 | 0.09 ± 1.05 | 1.27 ± 1.00 |
| Havens and Sigward 2015 | **Hip** | N | 90° | Mixed | -0.13 ± 0.85 | 0.37 ± 0.99 | 0.40 ± 0.70 |
| | | | | | | | |
| Havens and Sigward 2015 | **Knee** | N | 45° | Mixed | -4.15 ± 3.11 | | |
| Havens and Sigward 2015 | **Knee** | N | 90° | Mixed | -6.80 ± 2.16 | | |
| | | | | | | | |
| Havens and Sigward 2015 | **Ankle** | N | 45° | Mixed | -10.08 ±3.28 | | |
| Havens and Sigward 2015 | **Ankle** | N | 90° | Mixed | -6.87 ±2.91 | | |

Unanti = Unanticipated, N/Y = no/yes

**The running cut as a screening tool for Athletic Groin Pain**

Given the ecological validity of the running cut for field sport injuries, this screening test has been used to examine various sporting injuries. To date, two studies have examined the biomechanics of AGP using the running cut task. Edwards, Brooke and Cook, (2017) identified differences between those with and without AGP during this task in the transverse plane of the hip, knee and ankle. The second study, by our research group, investigated the biomechanical changes that occur with rehabilitation in a cohort of AGP patients (King et al. 2018). The authors identified multiple significant changes from pre- to post- rehabilitation across all joints examined. These findings highlight the potential usefulness of running cut as a screening exercise for AGP. Given the small number of subjects (n = 7) examined in Edwards, Brooke and Cook, (2017), along with the sole use of a pre- to post- rehabilitation study design used in King et al., (2018), the further examination of AGP biomechanics using this task is warranted.

### 2.7.2 The lateral hurdle hop

While the running cut is an ecologically valid means of assessing the mechanics of AGP in comparison to uninjured controls, one potential disadvantage of using very ecologically valid test actions is that they are typically characterised by a high level of variance. This variance, whilst representative, may mask any between group differences present. Conversely, while a very tightly controlled action will have good reliability and validity (e.g. isokinetic strength testing), if a test is too controlled, it may not replicate the demands of the actions associated with the injury. Clearly therefore, one should attempt to balance the external and internal validity of a test to maximise its ability to discern between injured and uninjured groups. An alternative to the running cut is the lateral hurdle hop test which aims to stress frontal plane control patterns and rate of force development similar to

the running cut, but in a more internally controlled manner. The following section will review the biomechanics of the lateral hurdle hop.

**Outline of the Lateral Hurdle Hop**

The participant begins on one foot, contralateral knee flexed at a 90-degree angle; hands are non-restricted and can be used for balance. The participant hops laterally over a (15cm) hurdle followed by an immediate hop back to the initial starting position as quickly as possible (Figure 2.9). Typical biomechanics for the first contact of the hurdle hop is presented in Figure 2.10 below. On landing the hip goes through an extension, flexion, extension pattern utilising a small range of motion. This is potentially utilised in an attempt to firstly resist eccentric loading and maintain an upright trunk position (Perry 1992) and then subsequently as a countermovement to help generate greater forces during propulsion. From 10-50% of ground contact the knee flexes through a moderate range of motion before it begins to extend. The ankle has a similar pattern as the knee as it rapidly dorsiflexes until approximately 50% of the ground contact and then subsequently plantarflexes.

As a test, single leg hops are commonly used as physical performance measures of function in terms of both rehabilitation (Myer et al. 2006) and as a screening tool in uninjured populations (Ostenberg and Roos 2000). Whilst a multitude of studies are evident investigating hopping and stiffness as demonstrated by a recent review (Lamontagne and Kennedy 2013) there is a dearth of literature investigating the kinematics and/or kinetics of frontal plane hopping. Three studies have investigated the biomechanics of a short lateral hop [as used within this current thesis] (Marshall et al. 2015, Monteleone et al. 2012, Yoshida, Taniguchi and Katayose 2011). In light of the lack of research into short lateral hopping movements, a similar movement pattern but over a longer distance as proposed by Fleischmann, Gehring, Mornieux, & Gollhofer, (2010) is also presented below.

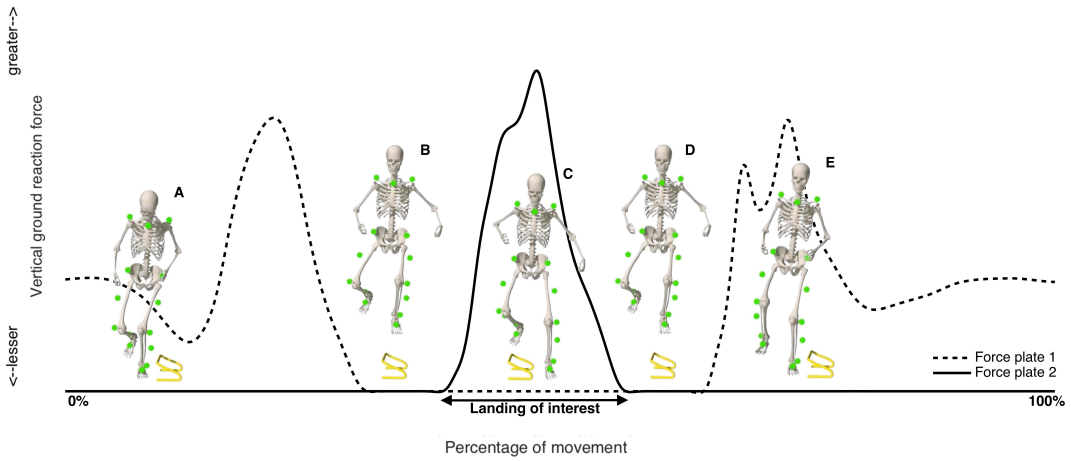


Figure 2.9: Graphical representation of the hurdle hop test. A) Starting position on force plate 1. B) Initial hop over hurdle. C) Initial landing phase that is biomechanically examined on force plate 2. D) Return hop back over the hurdle. E) End position on force plate 1 after hopping back over hurdle.

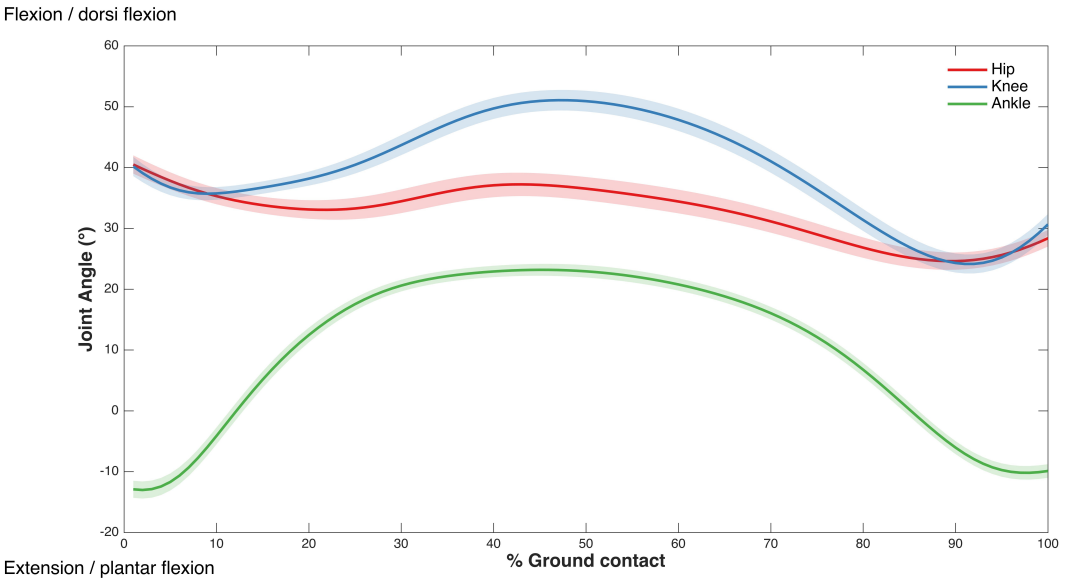


Figure 2.10: Sagittal plane kinematics of the hip, knee and ankle during a hurdle hop Shaded band indicates standard error.

The difference between the two movements is that with the short hop a relatively small horizontal displacement is utilized (approximately 40cm) in comparison to the long lateral hop in which the horizontal displacement has been reported between 96cm and 174cm. The following section will present both the kinematics and kinetics of the hurdle hop. Within the summary text when more than one study has presented the same variable, results will be presented as ranges. In

contrast when only study has presented results for a specific variable, the result presented will be the mean and standard deviation.

Kinematic results for the hurdle hop are presented in in Tables 2.14 to 2.17. At the thorax at initial contact, there is 15.1 $\pm$ 5.7° of trunk flexion, 2.0 $\pm$ 3.9° of contralateral side flexion and -2.4 $\pm$ 5.5° of ipsilateral rotation. The pelvis is anteriorly flexed 20.7 $\pm$ 5.0°, contralateral tilted -0.2 $\pm$ 4.3° and internally rotated 1.4 $\pm$ 5.5°. At the hip, there is between 29.9° to 34.8° of flexion, between 26.4° to 31.0° of abduction and -1.8 $\pm$ 7.2° of externally rotation. At the knee there is between 23.6° to 27.0° of flexion on initial contact, 1.1° to 1.8° of abduction and -3.5 $\pm$ 8.4° of internal rotation. At the ankle at initial contact there is between 12.1° and 13° of plantar-flexion, 0.7° and 12.4° of ankle inversion and -3.5 $\pm$ 4.4° of ankle adduction. At 140ms post initial contact, the hip flexes to 40.8 $\pm$ 5.8° and abducts 23.8 $\pm$ 8.1°. The knee flexes to 50.4 $\pm$ 3.9° and abducts to -0.3 $\pm$ 6.2°. At the ankle 17.5 $\pm$ 7.1° of ankle plantar-flexion is observed.

During the total eccentric phase, the trunk anteriorly flexed 6.8 $\pm$ 7.9° and contralateral side flexed 7.9 $\pm$ 5.9°. The pelvis is flexed 11.9 $\pm$ 4.4° and contralateral tilt of 1.4 $\pm$ 4.7°. The hip flexes to 34.0 $\pm$ 6.5° and abducts to 8.1 $\pm$ 5.3°. At the knee there is 42.3 $\pm$ 10.3° of flexion and 8.1 $\pm$ 5.3° of abduction. At the ankle there is 16.8 $\pm$ 4.2 of plantar-flexion and 4.5 $\pm$ 2.4° of ankle inversion. During the total contact phase for range of motion the ankle goes through 8.2 $\pm$ 1.2° of inversion.

Only one study from our research group has presented moments for the lateral hurdle hop test (Marshall et al. 2015). Moments are presented in Table 2.18 normalised to body mass.

Table 2.14: Angles (°) at initial contact for the lateral hurdle hop

| Author | Joint | Height (cm) | Distance (cm) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Weltin et al., 2014 | **Trunk** | NR | 100-126 cm | Mixed | 15.1 ± 5.7 | 2.0 ± 3.9 | −2.4 ± 5.5 |
| Weltin et al., 2014 | **Pelvis** | NR | 100-126 cm | Mixed | 20.7 ± 5.0 | −0.2 ± 4.3 | 1.4 ± 5.5 |
| Weltin et al., 2014 | **Hip** | NR | 100-126 cm | Mixed | 34.8 ± 6.9 | 31.0 ± 5.6 | −1.8 ± 7.2 |
| Mornieux et al 2014 | **Hip** | NR | 120cm | Mixed | 29.9 ± 6.5 | 26.4 ± 7.3 | |
| Weltin et al., 2014 | **Knee** | NR | 100-126 cm | Mixed | 27.0 ± 4.7 | 1.8 ± 5.0 | −3.5 ± 8.4 |
| Mornieux et al 2014 | **Knee** | NR | 120cm | Mixed | 23.6 ± 5.0 | 1.1 ± 3.9 | |
| Yoshida et al. 2011 | **Ankle** | NR | 120cm | Mixed | 12.1 ± 2.0 | 0.7 ± 4.2 | |
| Mornieux et al 2014 | **Ankle** | NR | 120cm | Mixed | 13 ± 8.9 | 12.4 ± 5.5 | |
| Monteleone et al. 2012 | **Ankle** | 14.3 | NR | Mixed | 12.1± 5.7 | 1.5 ± 5.1 | -3.5 ± 4.4 |

NR = not reported, Flexion/Dorsiflexion = +ive, Abduction/inversion/contra lateral tilt = +ive, Internal Rotation/ =+ive

Table 2.15: Angles (°) 140ms after initial contact for the lateral hurdle hop

| Author | Joint | Height (cm) | Distance (cm) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Mornieux et al 2014 | **Hip** | NR | 120cm | Mixed | 40.8 ± 5.8 | 23.8 ± 8.1 | |
| Mornieux et al 2014 | **Knee** | NR | 120cm | Mixed | 50.4 ± 3.9 | −0.3 ± 6.2 | |
| Mornieux et al 2014 | **Ankle** | NR | 120cm | Mixed | 17.5 ± 7.1 | | |

NR = not reported, Flexion/Dorsiflexion = +ive, Abduction/inversion/contra lateral tilt = +ive, Internal Rotation/ =+ive

Table 2.16: Angles (°) during the eccentric phase for the lateral hurdle hop

| Author | Joint | Height (cm) | Distance (cm) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Marshall et al., 2015 | **Trunk** | 15 cm | 40 cm | Male | 6.8 ± 7.9 | 7.9 ± 5.9 | |
| Marshall et al., 2015 | **Pelvis** | 15 cm | 40 cm | Male | 11.9 ± 4.4 | 1.4 ± 4.7 | |
| Marshall et al., 2015 | **Hip** | 15 cm | 40 cm | Male | 34.0 ± 6.5 | 8.1 ± 5.3 | |
| Marshall et al., 2015 | **Knee** | 15 cm | 40 cm | Male | 42.3 ± 10.3 | 3.1 ± 5.6 | |
| Marshall et al., 2015 | **Ankle** | 15 cm | 40 cm | Male | 16.8 ± 4.2 | 4.5 ± 2.4 | |

NR = not reported, Flexion/Dorsiflexion = +ive, Abduction/inversion/contra lateral tilt = +ive, Internal Rotation/ =+ive

Table 2.17: Joint ROM (°) during total contact for the hurdle hop

| Author | Joint | Height (cm) | Distance (cm) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Yoshida et al. 2011 | **Ankle** | NR | 30cm | Mixed | | 8.2 ± 1.2 | |

NR = not reported , Abduction/inversion/contra lateral tilt = +ive

Table 2.18: Joint moments (Nm/Kg) during the eccentric phase for the lateral hurdle hop

| Author | Joint | Height (cm) | Distance (cm) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Marshall et al., 2015 | **Hip** | 15 cm | 40 cm | Male | 2.9 ± 1.0 | -1.5 ± 0.3 | |
| Marshall et al., 2015 | **Knee** | 15 cm | 40 cm | Male | 2.6 ± 0.7 | -1.9 ± 0.6 | |
| Marshall et al., 2015 | **Ankle** | 15 cm | 40 cm | Male | 3.4 ± 0.5 | -0.4 ± 0.2 | |

Moments reported as internal moments. Extension/Plantarflexion = +ive, Abduction/inversion/contra lateral tilt = +ive, Internal Rotation/ =+ive

**The lateral hurdle hop as a screening tool for Athletic Groin Pain**

The lateral hurdle hop is a commonly utilized exercise to screen for anterior cruciate ligament ruptures and ankle sprains (Mornieux et al. 2014). Given that AGP is common to field sports that involve dynamic loading during multidirectional movements, the lateral hurdle hop exercise may also be an appropriate screening tool for this injury. Whilst poorly understood, some of the modifiable risk factors thought to be associated with AGP could influence control at the trunk, hip and pelvis (Janse van Rensburg et al. 2017, Edwards, Brooke and Cook 2017, King et al. 2018, Severin et al. 2017). To date no research has utilised the hurdle hop to explore AGP biomechanics. This may be a useful task to examine AGP as in comparison to the ecologically valid running cut test, the lateral hurdle hop may be a more internally controlled test, while retaining many of the dynamic loading features similar to the running cut.

### 2.7.3 The single leg drop landing

Given the lack of research investigating the biomechanics of the lateral hurdle hop, this section will also review the single leg drop landing (SLDL) given its similarities to the eccentric phase of the lateral hurdle hop landing. Landing, like its counterpart (jumping), is an integral part of most athletic activities. Injuries due to landing are common in field sports, such as Gaelic football, Hurling, Soccer and Australian football and account for between 4 and 15.9% of all injury incidences (Murphy et al. 2012; Blake et al. 2014; Hawkins 2001). Landing has been associated with numerous injuries such as anterior cruciate ligament injury (Kirkendall and Garrett 2000), ankle sprains (McKay 2001), patellar tendinopathy (Bisseling et al. 2007), patellar femoral pain syndrome (Boling et al. 2009) and stress fractures (Milner et al. 2006). For this reason landing biomechanics have been commonly analysed to identify risk of injury. To date, most investigations have examined

bilateral landing tasks e.g. (Huston et al. 2001, Kernozek et al. 2005). Despite this, research has reported that players landed unilaterally 67% and 41.3 - 48.8% of the time during netball (Lavipour 2011) and volleyball (Tillman et al., 2004; Lobietti and Coleman, 2010) respectively. Therefore, along with replicating the eccentric demands of the hurdle hop test, the single leg drop landing appears to be more ecologically valid than bilateral landing tasks. The following section will present the kinematics and kinetics of the single leg drop landing. Within the summary text when more than one study has presented the same variable, results will be presented as ranges. In contrast when only study has presented results for a specific variable, the result presented will be the mean $\pm$ standard deviation or conversely the median (range).

**Outline of the single leg drop landing**

The participant begins with one foot on the box, contralateral knee flexed at a 90-degree angle (Figure 2.11). The participant then drops off the box (ensuring minimal upward displacement) and lands on the force plate with the same foot that started on the box. On landing the hip, knee and angle flex as presented in Fig (Figure 2.12). The landing phase is defined between the events of initial contact and peak knee flexion.

The kinematics and kinetics of the single leg drop landing has been described under various conditions (e.g. different drop heights) and with various populations (e.g. different genders). For this reason, comparison of results between studies investigating single leg drop landings can be complicated. Of the 15 studies examined in this review, four reported a drop height of 20cm (Kiriyama, Sato and Takahira 2009, Ali, Robertson and Rouhi 2014, Lephart et al. 2002, Janse van Rensburg et al. 2017), five reported a drop height of 30cm (Nagano et al. 2007, Nagano et al. 2009, Orishimo et al. 2009, Schmitz et al. 2007, Vairo et al. 2008), one study reported a drop height of 33cm (Gardner et al. 2012), two studies

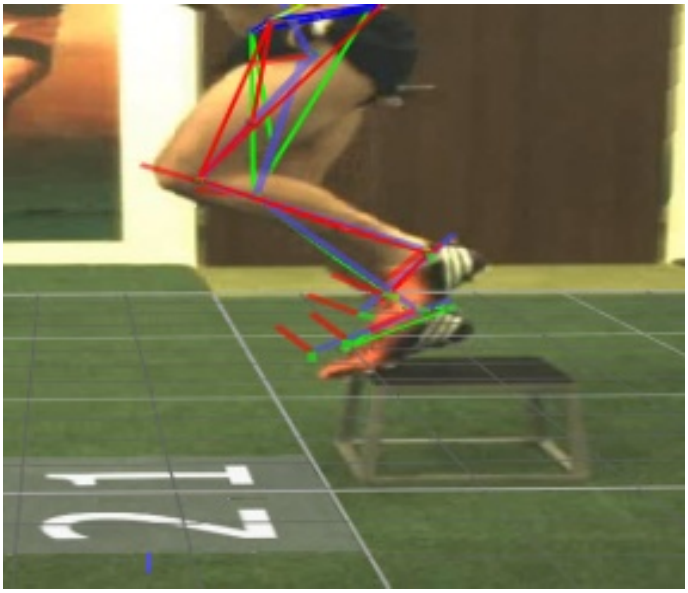

Figure 2.11: Single leg drop landing

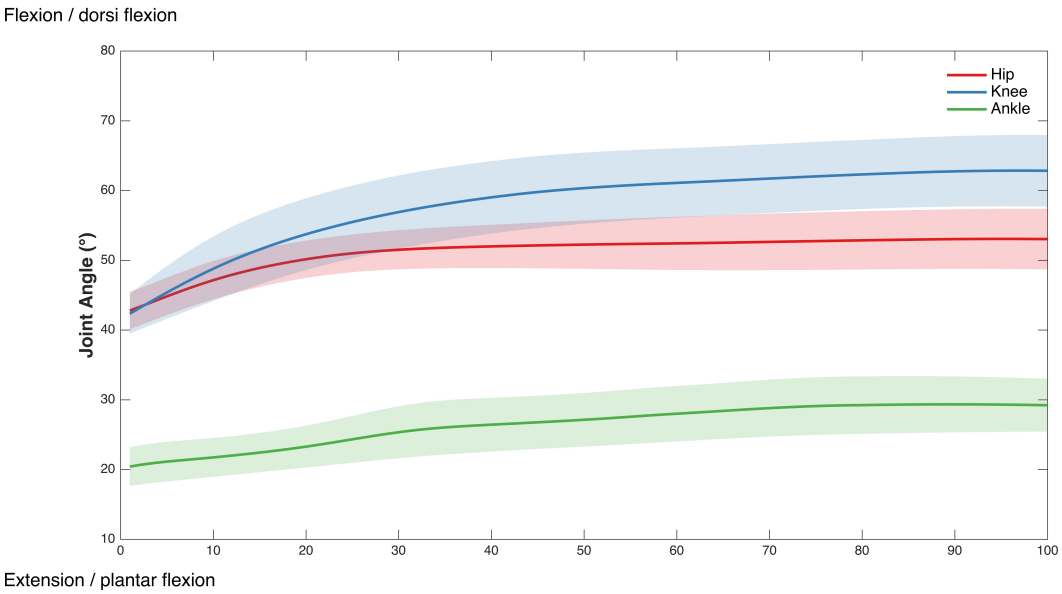


Figure 2.12: Sagittal plane kinematics of the hip, knee and ankle during a hurdle hop Shaded band indicates standard error.

reported a drop height of 40cm (Ali, Robertson and Rouhi 2014, Pappas et al. 2007), four studies examined a drop height from 60cm (Russell et al. 2006, Ali, Robertson and Rouhi 2014, Yeow, Lee and Goh 2011, Garrison and Hart 2005) and one study reported drop height at the maximum vertical jump height of the subject (Weinhandl, Joshi and O'Connor 2010). In the case of (Weinhandl, Joshi and O'Connor 2010) this was a mean value of 44cm and 28.2cm for males and females, respectively.

Full kinematic results are presented in in Table 2.19 for initial contact. On landing at initial contact the trunk is flexed to angles of between 12.7° and 16.4° of flexion. At the pelvis values of 6.78 (range: 37.01°), -10.56 (range: 25.47°) and 6.14 (range: 20.06°) are reported for anterior tilt, ipsilateral drop and internal rotation, respectively. The hip is flexed to between 15.7 and 30.0°, adducted between -7.29 and -8.2° and internally rotated to 3.14 (48.79°). At the knee angles of between 12.8° and 42.5° of flexion, between -4.0° and 6.8° of abduction and between 1.2° and 13.7° of internal rotation are recorded. At the ankle between -25.8 ° and -0.22 ° of plantar flexion and 0.9 $\pm$ 0.8° of abduction are observed.

Full kinematic results are presented in Table 2.20 for ROM during the landing phase. During the landing phase at the pelvis, 4.41(14.45°) of motion are recorded in the sagittal plane, 10.56 (25.47°) in the frontal plane and 6.14(20.06°) in the transverse plane. The hip moves through 1.6° to 25.4° of motion in the sagittal plane, in the frontal plane between 6.09 ° and 16.2°, and between 3.08 and 3.14° in the transverse plane. At the knee 8.3° to 60.3° of motion are reported for the sagittal plane, whilst between 8.5° and 11.3° of motion is recorded for the frontal plane. At the ankle between 26.7° and 48.0° of motion are recorded for the sagittal plane.

In respect to athletic groin pain, it has been suggested that hip and pelvic control particularly in the transverse and frontal planes may be of importance to its development (Garvey et al., 2010). Additionally as noted by (Prior et al. 2014)

Table 2.19: Joint Angle (°) at initial contact during single leg drop landing

| Author | Joint | Height (cm) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|
| Ali et al., 2014 | **Trunk** | 20 cm | Males | 13.9 ± 3.2 | . | . |
| Ali et al., 2014 | **Trunk** | 40 cm | Males | 12.74 ± 2.1 | . | . |
| Ali et al., 2014 | **Trunk** | 60 cm | Males | 16.4 ± 3.1 | . | . |
| | | | | | | |
| Janse van Rensburg et al., 2017 | **Pelvis** | 20 cm | Male | 6.78 ( 37.01) | -10.56 (25.47) | 6.14 (20.06) |
| | | | | | | |
| Orishimo *et al.,* 2009 | **Hip** | 30 cm | Male | -2.6 ± 10.7 | -10.3 ± 5.1 | . |
| Orishimo *et al.,* 2009 | **Hip** | 30 cm | Female | 5.9 ± 8.5 | -12.5 ± 5.0 | . |
| Yeow *et al.,* 2011 | **Hip** | 60 cm | Male | 30.0 ± 4.4 | - 8.2 ± 4.8 | . |
| Ali et al., 2014 | **Hip** | 20 cm | Male | 23.3 ± 7.3 | . | . |
| Ali et al., 2014 | **Hip** | 40 cm | Male | 19.02 ± 5.2 | . | . |
| Ali et al., 2014 | **Hip** | 60 cm | Male | 21.04 ± 6.9 | . | . |
| Weinhandl *et al.,* 2010 | **Hip** | Max Jump (44cm) | Male | 16.8 ± 8.6 | . | . |
| Weinhandl *et al.,* 2010 | **Hip** | Max Jump (28.2cm) | Female | 15.7 ± 9.3 | . | . |
| Schmitz *et al.,* 2007 | **Hip** | 30 cm | Male | 16.7 ± 7.6 | . | . |
| Schmitz *et al.,* 2007 | **Hip** | 30 cm | Female | 21.6 ± 6.3 | . | . |
| Vairo *et al.,* 2008 | **Hip** | 30 cm | Mixed | 23.6 ± 6.58 | . | . |
| Janse van Rensburg et al., 2017 | **Hip** | 20 cm | Male | 24.65 (62.84) | -9.77 (30.03) | 3.14 (48.79) |
| | | | | | | |
| Ali et al., 2014 | **Knee** | 20cm | Males | 30.8 ±5.9 | . | . |
| Ali et al., 2014 | **Knee** | 40 cm | Males | 32.9 ±3.0 | . | . |
| Ali et al., 2014 | **Knee** | 60 cm | Males | 35.89 ±2.3 | . | . |
| Kiriyama *et al.,* 2009 | **Knee** | 20 cm | Males | . | . | 10.1 ± 5.5 * |
| Kiriyama *et al.,* 2009 | **Knee** | 20 cm | Females | . | . | 13.7 ± 9.1* |
| Nagano *et al.,* 2007 | **Knee** | 30 cm | Males | 15.9 ±1.5 | 2.0 ± 0.7 | 1.2 ± 1.4 * |
| Nagano *et al.,* 2007 | **Knee** | 30 cm | Females | 18.0 ±1.5 | 1.8 ± 0.6 | 2.2 ± 1.4 * |
| Nagano *et al.,* 2009 | **Knee** | 30 cm | Females | 15.8 ± 5.0 | -4.0 ± 2.0 | 9.0 ± 3.4 * |
| Orishimo *et al.,* 2009 | **Knee** | 30 cm | Males | 1.0 ± 7.0 | 0.03 ± 2.5 | . |
| Orishimo *et al.,* 2009 | **Knee** | 30 cm | Females | 3.5 ± 4.4 | -1.3 ± 3.7 | . |
| Schmitz *et al.,* 2007 | **Knee** | 30 cm | Males | 38.9 ± 7.1 | . | . |
| Schmitz *et al.,* 2007 | **Knee** | 30 cm | Females | 42.5 ± 9.4 | . | . |
| Russel *et al.,* 2006 | **Knee** | 60 cm | Males | . | 3.85 ± 4.03 | . |
| Russel *et al.,* 2006 | **Knee** | 60 cm | Females | . | -0.65 ±3.32 | . |
| Yeow *et al.,* 2011 | **Knee** | 60 cm | Males | 20.8 ± 5.2 | 6.8 ± 3.9 | . |
| Weinhandl *et al.,* 2010 | **Knee** | Max Jump (44cm) | Males | 12.8 ± 5.8 | . | . |

Table 2.19: Joint Angle (°) at initial contact during single leg drop landing (cont.)

| | | | | | | |
|---|---|---|---|---|---|---|
| Weinhandl *et al.*, 2010 | **Knee** | Max Jump (28.2cm) | Females | 12.8 ± 5.6 | . | . |
| Pappas *et al.*, 2007 | **Knee** | 40 cm | Mixed | 15.1 (7.7) | . | . |
| Vairo *et al.*, 2008 | **Knee** | 30 cm | Mixed | 18.8 ± 7.72 | . | . |
| | | | | | | |
| Orishimo *et al.*, 2009 | **Ankle** | 30 cm | Males | -33.2 ± 5.0 | -12.9 ±10.0 | . |
| Orishimo *et al.*, 2009 | **Ankle** | 30 cm | Females | -34.2 ± 4.6 | -6.9 ± 7.5 | . |
| Schmitz *et al.*, 2007 | **Ankle** | 30 cm | Males | -24.3 ± 6.3 † | . | . |
| Schmitz *et al.*, 2007 | **Ankle** | 30 cm | Females | -25.0 ± 5.6 † | . | . |
| Ali et al., 2014 | **Ankle** | 20 cm | Males | -0.22 ± 3.52 | . | . |
| Ali et al., 2014 | **Ankle** | 40 cm | Males | -2.11 ± 2.44 | . | . |
| Ali et al., 2014 | **Ankle** | 60 cm | Males | -2.67 ± 3.54 | . | . |
| Yeow *et al.*, 2011 | **Ankle** | 60 cm | Males | -17.3 ± 6.9 | 0.9 ± 0.8 | . |
| Weinhandl *et al.*, 2010 | **Ankle** | Max Jump (44cm) | Males | -25.8 ± 9.7 | . | . |
| Weinhandl *et al.*, 2010 | **Ankle** | Max Jump (28.2cm) | Females | -23.1 ± 10.3 | . | . |
| Vairo *et al.*, 2008 | **Ankle** | 30 cm | Mixed | -18.7 ± 7.26 | . | . |

Flexion/Dorsiflexion = +ive, Abduction = +ive, Internal Rotation =+ive, '*' = tibia rotation, '†' = published value transformed by 90°

trunk and/or pelvic positioning during a single leg stance can result in a large change to hip and thigh muscle activation which may predispose to muscular overload.

Table 2.20: Joint ROM (°) during single leg drop landing

| Author | Joint | Height (cm) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|
| Janse van Rensburg et al., 2017 | **Pelvis** | 20 cm | Male | 4.41 (14.45) | 10.56 (25.47) | 6.14 (20.06) |
| Weinhandl *et al.*, 2010 | **Hip** | Max Jump (44cm) | Male | 25.4 ± 7.4 | | |
| Weinhandl *et al.*, 2010 | **Hip** | Max Jump (28.2cm) | Female | 16.8 ± 4.7 | | |
| Schmitz *et al.*, 2007 | **Hip** | 30 cm | Male | 4.0 ± 4.4 | | |
| Schmitz *et al.*, 2007 | **Hip** | 30 cm | Female | 1.6 ± 1.9 | | |
| Orishimo *et al.*, 2009 | **Hip** | 30 cm | Male | 25.4 ± 7.4 | 15.5 ± 4.2 | |
| Orishimo *et al.*, 2009 | **Hip** | 30 cm | Female | 24.3 ± 5.8 | 16.2 ± 5.1 | |
| Lephart *et al.*, 2002 | **Hip** | 20 cm | Male | 6.65 ± 4.91 | 6.09 ± 3.53 | 3.08 ± 2.16 |
| Lephart *et al.*, 2002 | **Hip** | 20 cm | Female | 7.12 ± 5.57 | 10.67 ± 8.85 | 7.49 ± 3.69 |
| Janse van Rensburg et al., 2017 | **Hip** | 20 cm | Male | 9.93 (25.69) | 9.77 (30.03) | 3.14 (48.8) |
| Weinhandl *et al.*, 2010 | **Knee** | Max Jump (44cm) | Male | 49.8 ± 9.7 | | |
| Weinhandl *et al.*, 2010 | **Knee** | Max Jump (28.2cm) | Female | 41.6 ± 7.0 | | |
| Schmitz *et al.*, 2007 | **Knee** | 30 cm | Male | 12.9 ± 6.9 | | |
| Schmitz *et al.*, 2007 | **Knee** | 30 cm | Female | 8.3 ± 5.9 | | |
| Orishimo *et al.*, 2009 | **Knee** | 30 cm | Male | 60.3 ± 6.8 | 8.5 ± 4.4 | |
| Orishimo *et al.*, 2009 | **Knee** | 30 cm | Female | 56.3 ± 4.8 | 11.3 ± 5.8 | |
| Lephart *et al.*, 2002 | **Knee** | 20 cm | Male | 31.10 ± 9.92 | | |
| Lephart *et al.*, 2002 | **Knee** | 20 cm | Female | 17.41 ± 12.96 | | |
| Weinhandl *et al.*, 2010 | **Ankle** | Max Jump (44cm) | Male | 48.0 ± 8.3 | | |
| Weinhandl *et al.*, 2010 | **Ankle** | Max Jump (28.2cm) | Female | 43.1 ± 11.7 | | |
| Schmitz *et al.*, 2007 | **Ankle** | 30 cm | Male | 26.7 ± 5.7 | | |
| Schmitz *et al.*, 2007 | **Ankle** | 30 cm | Female | 27.1 ± 6.0 | | |

Flexion/Dorsiflexion = +ive, Abduction = +ive, Internal Rotation =+ive

The joint moments of the single leg landing has been poorly described in the literature. Only two studies (Weinhandl, Joshi and O'Connor 2010, Garrison and Hart 2005) examined the moments of this movement. Moments are presented in Table 2.21 normalised to body mass (Garrison and Hart 2005) and body mass times the square root of the landing height (Weinhandl, Joshi and O'Connor 2010). In addition to the lack of literature in general, it can be noted that in particular

Table 2.21: Peak joint moments (Nm/Kg) during single leg drop landing

| Author | Joint | Height (cm) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|
| Weinhandl *et al.*, 2010 | **Hip** | Max Jump (44cm) | Males | -2.9 ± 1.2 | -3.0 ± 0.8 | . |
| Weinhandl *et al.*, 2010 | **Hip** | Max Jump (28.2cm) | Females | -3.0 ± 1.6 | -3.4 ± 0.7 | . |
| | | | | | | |
| Garrison et al., 2005 | **Knee** | 60 cm | Males | -1.27 ± 0.90 | 2.2 ± 0.46 | 0.40 ± 0.18 |
| Garrison et al., 2005 | **Knee** | 60 cm | Females | -1.06 ± 0.3 | 1.67 ± 0.34 | 0.49 ± 0.06 |
| Weinhandl *et al.*, 2010 | **Knee** | Max Jump (44cm) | Males | -3.6 ± 1.0 | 0.3 ± 0.3 | . |
| Weinhandl *et al.*, 2010 | **Knee** | Max Jump (28.2cm) | Females | -3.4 ± 1.2 | 0.4 ± 0.4 | . |
| | | | | | | |
| Weinhandl *et al.*, 2010 | **Ankle** | Max Jump (44cm) | Males | -3.4 ± 1.3 | -0.5 ± 0.2 | . |
| Weinhandl *et al.*, 2010 | **Ankle** | Max Jump (28.2cm) | Females | -3.3 ± 1.5 | -0.6 ± 0.4 | . |

Moments reported as internal moments. Flexion/dorsiflexion =+ive, Adduction/inversion = +ive,

there is little data reported on transverse plane moments. This may be particularly important at the hip for athletic groin pain, where knowledge of kinetics can help understand control of the femur on pelvis rotations.

Energy dissipating strategies are important in landing. Despite this, of the 19 papers included in this review only five papers reported the energetics, one of which (Schmitz et al. 2007) was excluded due to unusual normalisation. Full results are presented in Table 2.22 and Table 2.23 below. Of the four remaining papers only one study (Ali, Robertson and Rouhi 2014) reported joint power whilst the rest reported only joint work. In the sagittal plane the ankle contributes the most to energy dissipation with between 42 - 47% of the total work dissipated. The contribution of the hip and knee vary between studies, which may be due to the different heights used. There also seems to be a trend for females to use a more knee dominant strategy compared to males (Weinhandl, Joshi and O'Connor 2010, Yeow, Lee and Goh 2011, Gardner et al. 2012). There is no literature that

Table 2.22: Joint work (J/Kg) during single leg drop landing

| Author | Joint | Type | Height (cm) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|---|
| Yeow *et al.,* 2011 | **Hip** | Peak | 60 cm | Males | -2.07 ± 0.72 | -0.14 ± 0.12 | . |
| Weinhandl *et al.,* 2010 | **Hip** | Net | Max Jump (44cm) | Males | -1.40 ± 0.98† | . | . |
| Weinhandl *et al.,* 2010 | **Hip** | Net | Max Jump (28.2cm) | Females | -1.18 ± 0.54† | . | . |
| Gardener *et al.,* 2012 | **Hip** | Net | 33cm | Females | -0.13 ± 0.12 | . | . |
| | | | | | | | |
| Yeow *et al.,* 2011 | **Knee** | Peak | 60 cm | Males | -0.55 ± 0.28 | -0.23 ± 0.09 | . |
| Weinhandl *et al.,* 2010 | **Knee** | Net | Max Jump (44cm) | Males | -2.95 ± 0.97† | . | . |
| Weinhandl *et al.,* 2010 | **Knee** | Net | Max Jump (28.2cm) | Females | -2.77 ± 0.91† | . | . |
| Gardener *et al.,* 2012 | **Knee** | Net | 33 cm | Females | -1.61 ± 0.45 | . | . |
| Ali et al., 2014 | **Knee** | Net* | 20cm | Male | −0.86±0.36 | . | . |
| Ali et al., 2014 | **Knee** | Net* | 40 cm | Male | −1.33±0.44 | . | . |
| Ali et al., 2014 | **Knee** | Net* | 60 cm | Male | −1.89±0.56 | . | . |
| | | | | | | | |
| Yeow *et al.,* 2011 | **Ankle** | Peak | 60 cm | Males | -2.20 ± 0.30 | - 0.01 ± 0.01 | . |
| Weinhandl *et al.,* 2010 | **Ankle** | Net | Max Jump (44cm) | Males | -3.15 ± 0.78 † | . | . |
| Weinhandl *et al.,* 2010 | **Ankle** | Net | Max Jump (28.2cm) | Females | -3.54 ± 1.19 † | . | . |
| Gardener *et al.,* 2012 | **Ankle** | Net | 33cm | Females | -1.28 ± 0.12 | . | . |

′†′ = Exact values obtained through personal correspondence with author. * = Net of eccentric power only

Table 2.23: Joint power (W/kg) during single leg drop landing

| Author | Joint | Height (cm) | Gender | Sagittal | Frontal | Transverse |
|---|---|---|---|---|---|---|
| Ali et al., 2014 | **Knee** | 20cm | Male | −5.05 ± 2.95 | | |
| | **Knee** | 40 cm | Male | −10.77 ± 3.03 | | |
| | **Knee** | 60 cm | Male | −15.54 ± 4.40 | | |

reports transverse plane work, and only one study (Yeow, Lee and Goh 2011) investigating frontal plane work. For athletic groin pain these variables may be important, since non-sagittal energy dissipation will likely affect how the muscle, tendons and bony structures of the anterior pelvis absorbs force during loading.

**The single leg drop landing as a screening tool for Athletic Groin Pain**

To date only one study has examined the biomechanics of AGP during a SLDL (Janse van Rensburg et al., 2017). The authors investigated hip and pelvis kinematics in 10 AGP patients in comparison to 10 controls and identified differences between the two groups in both the frontal and transverse plane of the hip and pelvis. Despite this, within this current thesis, the SLDL test was not utilised to examine the biomechanics associated with AGP. AGP is common is sports involving rapid twisting and turning (Thorborg et al. 2017, Orchard, Seward and Orchard 2013, Werner et al. 2009), as such it was hypothesised that test actions involving multiple planes (such as the running cut and lateral hurdle hop) would more closely replicate the demands of field based sport actions.

### 2.7.4 Summary

This chapter section presented uninjured movement mechanics for the two test actions utilised within this thesis (the running cut and lateral hurdle hop), in addition to the biomechanics of the single leg drop landing. While it is difficult to compare across studies and tests, it would appear that the hurdle hop and cutting task are more similar to each other in comparison to the single leg drop landing. This may be explained by the use of both a lateral movement and the stretch shortening cycle in the hurdle hop test, which is absent in the single leg drop landing. Despite some similarities however, the running cut and lateral hurdle hop are characterised by distinct biomechanical features and will likely stress the mechanics of AGP patients in different ways. For example, during the running cut, the thorax demonstrates much larger angles in comparison to the lateral hurdle hop. Conversely, the hurdle hop is characterised by much larger moments at the ankle joint when compared to running cut task.

While injuries due to landing are common in field sports (Hawkins, 2001; Murphy et al., 2012; Blake et al., 2014), given that AGP is typically associated with agility type tasks (Werner et al., 2009; Orchard, Seward and Orchard, 2013; Thorborg et al., 2017), the single leg drop landing was not utilised within this thesis as a testing action.

The running cut was utilised as an ecological examination of AGP and was conducted in a planned manner. While an unanticipated task would have higher ecological validity and loading than a planned running cut (Besier et al., 2001), the potential for greater variance in the unanticipated task itself could result in masking of between group differences and as such was not examined. Similarly, while the hurdle hop task shares some biomechanical similarities to a side stepping/ running cut test, it has the potential advantage being more internally controlled than the running cut test. Examination of AGP during both a lateral hurdle hop and running cut task is therefore warranted to explore the biomechanics related to this injury.

## 2.8 Subgroup Analysis

Research typically examines the biomechanical risk factors for an injury using a single group study design. Such an approach assumes that the single group examined is suitably homogeneous in nature, that is, there is no underlying participant characteristic that may confound the study's findings. For example when investigating the risk factors for ACL injury, researchers commonly delimit their study to one sex as it is accepted that males and females may demonstrate distinct risk factors for this injury (Sigward and Powers, 2006; Nagano et al., 2007; Weinhandl, 2007; Beaulieu, Lamontagne and Xu, 2008; Weinhandl, Joshi and O'Connor, 2010; Weltin, Gollhofer and Mornieux, 2015; Schreurs, Benjaminse and Lemmink, 2017). Similarly, researchers interested in the performance determining

factors of a sport may choose to delimit their research to a specific age group as the morphological and physiological changes that occur with age may confound the performance determining factors identified (Shephard, 1992; Maharam et al., 1999; Diallo et al., 2001; Dor é et al., 2001; Tanaka and Seals, 2008; Ransdell, Vener and Huberty, 2009; Reaburn and Dascombe, 2009; Faigenbaum and Myer, 2010). This approach of delimiting the study design is logical and can allow for the identification and targeting of more specific risk factors for each of the populations examined. However, a potential limitation to this approach is the requirement for a priori knowledge of these potentially confounding factors. An alternative method is to utilise a statistical subgrouping technique known as clustering to identify similar subgroups within a larger sample/population (Rein et al., 2010). Clustering is known as an unsupervised technique as it separates and organizes unlabelled data into different groups whose members are similar to each other in some metric (Segaran, 2007). The benefit of utilising clustering is that it does not rely on the researcher's prior knowledge to identify subgroups. For example, when examining the performance determining factors of counter movement jumping, researchers have appropriately delimited their study for potentially confounding factors such as age and sex (Hubley and Wells, 1983; Bobbert et al., 1986; Aragon-Vargas and Melissa Gross, 1997; Vanezis and Lees, 2005; Vanrenterghem, Lees and Clercq, 2008). Despite this, there remained contrasting findings in the literature with regards to the performance related factors identified for this task. Later research identified the presence of four biomechanical subgroups within a cohort of 122 male athletes (Richter et al., 2014). Not only were the authors able to provide a greater ability to describe jump height in comparison to a whole group analysis, but the authors also identified that each cluster was characterised by its own performance determining factors. Hence, contrasting findings between previous studies that examined vertical jumping at a whole group level may be explained, at least in part, by the false assumption that a group of male athletes of

similar age and experience would utilise the same mechanics to achieve the task of jumping. Similarly, Phinyomark et al., (2015) identified two distinct running gait patterns in a group of healthy runners independent of age, height, weight, and running speed. When these two groups were separately compared to a large cohort of runners experiencing patellofemoral pain, two different risk factors were identified in relation to knee abduction angle. This again highlights the potential benefit of using subgroup over whole group study designs.

Clustering has been utilised successfully in various areas such as identifying performance determining factors (Richter et al., 2014), recognising pathological gaits (Toro, Nester and Farren, 2007; Roche et al., 2014) and identifying risk factors for injuries (Phinyomark et al., 2015). To date however, only one study by our research group (Franklyn-Miller et al., 2017) has examined clustering with respect to AGP. The authors examined 322 athletes with AGP during a 110° cutting task and utilised a clustering technique to identify three distinct subgroups within the cohort. Cluster 1 (40%) was characterised by increased ankle eversion, external rotation and knee internal rotation and greater knee work in comparison to the other two Clusters. Cluster 2 (15%) was characterised by increased hip flexion, pelvis contralateral drop, thorax tilt and increased hip work in comparison to the other two Clusters. Cluster 3 (45%) was characterised by high ankle dorsiflexion, thorax contralateral drop, ankle work and prolonged ground contact time in comparison to the other two Clusters. Interestingly, the authors demonstrated no association between movement cluster and original clinical diagnosis, suggesting that site of pain may be of less importance than the possible propagative mechanisms. These findings support the concept of examining the biomechanics of AGP as a single clinical entity but also suggests that the biomechanical examination of AGP may benefit from the use of clustering. In light of these findings the Franklyn-Miller et al., (2017) suggest that the identified

movement clusters may represent targets for rehabilitation, however this requires further investigation.

### 2.8.1 Clustering Techniques

There are various different forms of clustering utilised within the literature. While a review of all methods is outside the scope of this current thesis, this section will provide a brief overview of two of the most commonly utilised approaches namely K-means clusters and Hierarchal clustering.

#### K means clustering

In k-means clustering, the number of clusters is chosen prior to analysis and individuals are assigned into groups based on the location of their observations (Martinez and Martinez, 2004). Given a data set (Figure 2.13.A), K-means clustering starts with k randomly placed centroids and each individual is assigned to the nearest centroid (Figure 2.13.B). After the initial assignment, each centroid location is optimised by is moving the centroid to the average location of its members and the individual data points are again reassigned to the nearest centroid (Figure 2.13.C). This process of assignment and optimisation is repeated until the members of a centroid stop changing (Figure 2.13.D)

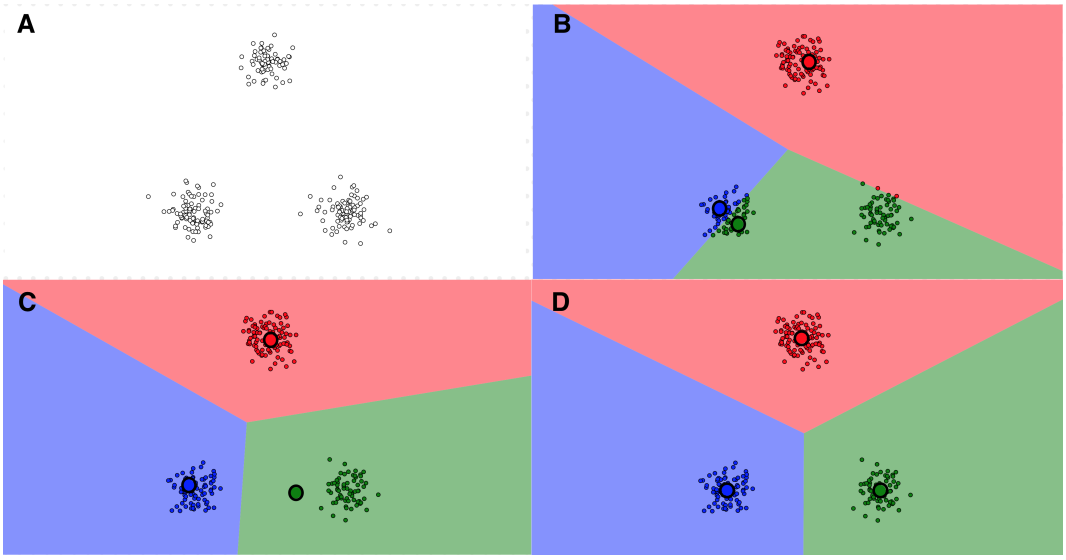

Figure 2.13: Process of K - Means clustering with 3 clusters

One disadvantage of K - Means clustering is the requirement to choose the number of clusters prior to analysis (Segaran, 2007). This is problematic in biomechanics as it is often difficult to determine a priori how many clusters will be present within a larger cohort. An alternative popular method is hierarchical clustering which will be utilized in this thesis.

### Hierarchical Clustering

In hierarchical clustering, one does not have to know the number of groups ahead of time and as such, this approach is commonly used in exploratory research (Segaran, 2007). Hierarchical clustering consists of a sequence of steps, where two groups are either merged (agglomerative) or divided (divisive) (Martinez and Martinez, 2004). Both these approaches are very similar, but the process of clustering is opposite to one another. Agglomerative approaches are more commonly used and as such will only be covered in this thesis. Figure 2.14, illustrates the Hierarchical clustering process (Segaran, 2007). At first all the individuals' data points start as individual clusters. Subsequently, it calculates the distance between every individual and searches for the two individuals that

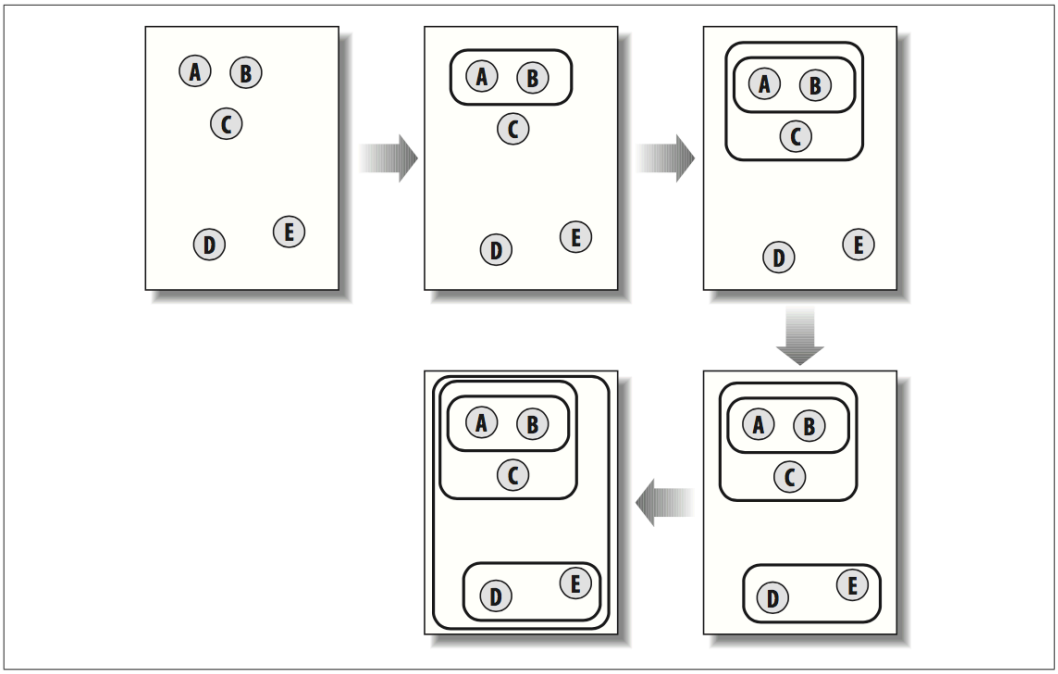


Figure 2.14: Process of Hierarchical Clustering (Segaran, 2007)

are most similar to one another (based on a distance metric) to create a new cluster. For example in Figure 2.14, within the second step, A and B, the two items closest together have merged to form a new cluster. The hierarchical clustering algorithm repeats this process until it has created one single group that contains every individual (Segaran, 2007).

After the hierarchical clustering algorithm is complete, it is common to visualise the findings in a graph termed a dendrogram (Figure 2.15). This dendrogram can then be explored to help understand how the clusters were formed and along with various gap statistics (Martinez and Martinez, 2004) can help choose the number of clusters to retain (Segaran, 2007).

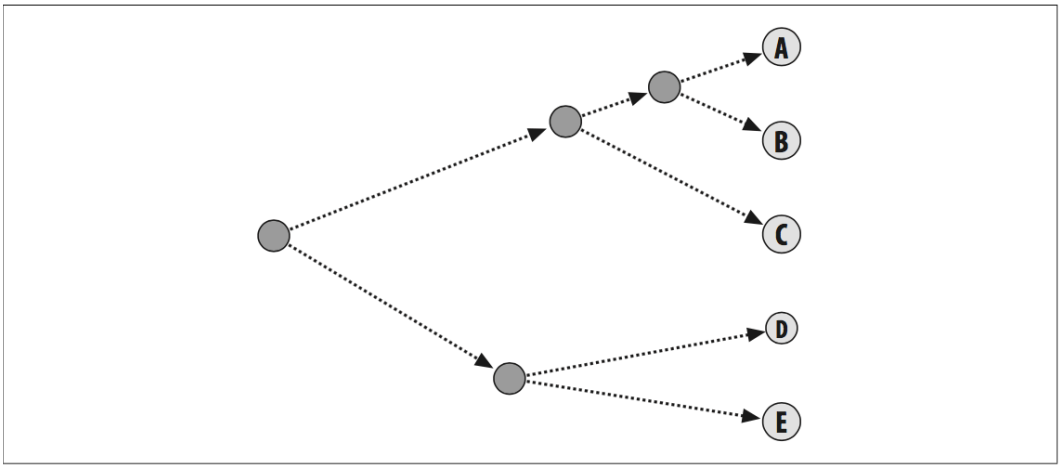

Figure 2.15: Dendrogram utilized to visualize the hierarchical clustering process (Segaran, 2007)

### 2.8.2 Summary

Sub-group analysis using a clustering technique can be a useful means of exploring the existence of sub-groups within a larger population. To date only one study by our research group (Franklyn-Miller et al., 2017) has examined clustering with respect to AGP. Future research is therefore warranted to further explore the use of cluster in AGP to further our understanding of this injury.

## 2.9 Discrete Point vs. Continuous Waveform Analysis

Typically, biomechanical data characterising human movement is collected as a temporal waveform or time series (e.g. joint angles). This biomechanical data has traditionally been examined using discrete point analysis, whereby the total waveform is represented by a pre-selected summary value(s) (e.g. maximum or mean value). While this approach has been successfully utilised to compare groups (Kernozek et al., 2005; Russell et al., 2006; Weinhandl, Joshi and O'Connor, 2010; Sinclair et al., 2015; Edwards, Brooke and Cook, 2017; Janse van Rensburg et al., 2017) discrete point analysis is subject to three key limitations. Firstly, discrete

point analysis requires the researcher to pre-select a key metric (discrete point) to statistically examine. This however relies on the experience of the researcher and introduces a biased assumption regarding the spatio-temporal focus within the waveform. In fact, selecting various different points to examine has been suggested as a reason for inconsistent conclusions within the literature (O'Connor and Bottum, 2009). Secondarily, discrete point analysis does not consider the vast majority of the signal available. For example when examining risk factors for anterior cruciate ligament, researchers have focused on the knee abductor moment (Lin et al., 2012; Sigward, Pollard and Powers, 2012; Schreurs, Benjaminse and Lemmink, 2017). Assuming a biomechanical waveform normalized to 101 data point (0-100% of the action examined), the examination of the single discrete point within the knee abductor moment would result in discarding almost 99% of the data available. Finally, the extraction of discrete points to statistically analyse, hazards the possibility of comparing metrics that are not functionally similar should they occur at different temporal stages of a movement. For example Figure 2.16 depicts unimodal (Figure 2.16.A) and bimodal (Figure 2.16.B) variants of a ground reaction curve during a hurdle hop test. The examination of the peak ground reaction forces from these two profiles may result in erroneous findings.

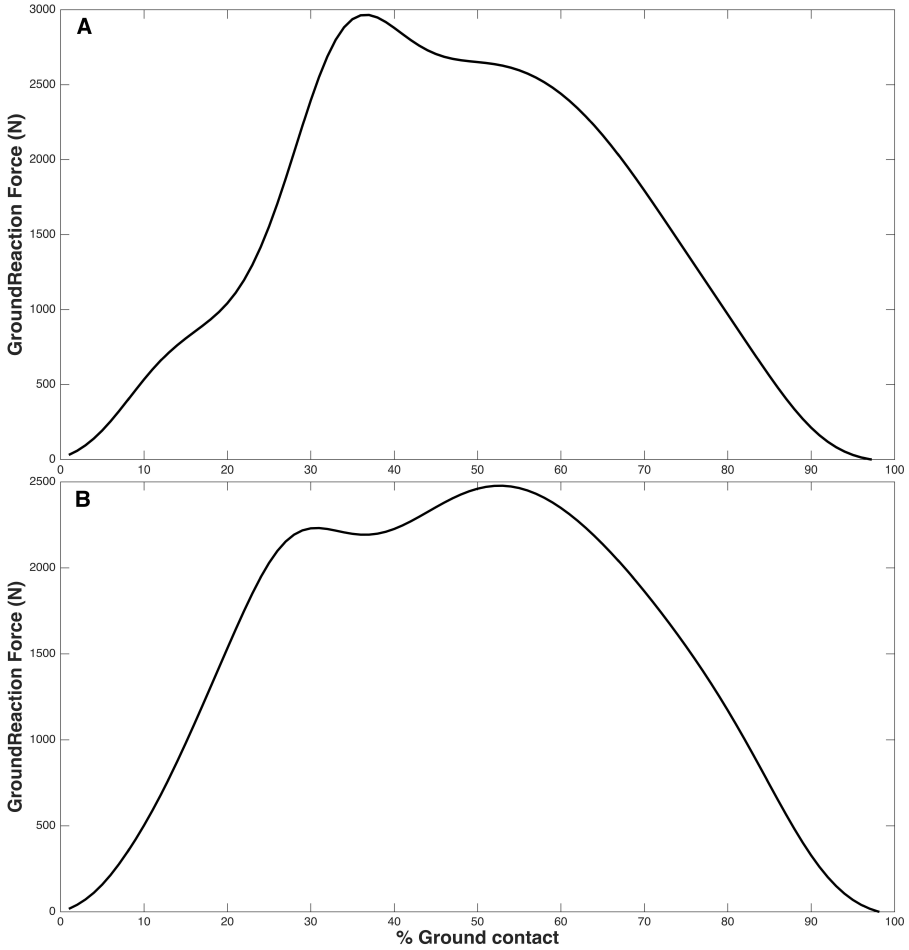

Figure 2.16: Illustration of Unimodal and Bimodal Waveforms

Considering the above limitations associated with discrete point analysis, in recent years, researchers have begun to explore alternative techniques of examining the whole waveform or key phases. This has been done in numerous ways, for example using principal component analysis (Federolf, Boyer and Andriacchi, 2013), functional data analysis (Ramsay and Silverman, 2005), analysis of characterizing phases (Richter, O'Connor, et al., 2014b) and statistical parametric mapping (Pataky, 2010). The benefits of continuous waveform methods have been observed in its enhanced ability to explain jump height (Richter, O'Connor, et al., 2014b), identify asymmetries (Marshall et al., 2015) and detect differences between injured subjects and healthy controls (Donoghue et al., 2008).

While an in-depth critic of the various continuous waveform methods available is outside the scope of this current thesis, the interested reader is referred to (Chau, 2001a, 2001b) for a review of prominent continuous waveform methods utilized within the biomechanics literature. Further, Appendix I of this thesis contains a brief overview of the two continuous waveform analysis methods utilized in this thesis, namely, analysis of characterizing phases (Richter, O'Connor, et al., 2014b) and statistical parametric mapping (Pataky, 2010).

## 2.10 Literature Review Summary

Given the high prevalence of AGP during sports involving repetitive agility tasks, the biomechanical evaluation of both the technique and joint loading that occurs during these tests is an essential step to enhance our understanding of this injury. Whilst this has been conducted quite extensively with other injuries, until 2017 there was no published research investigating the 3D biomechanical risk factors of AGP.

Due to the paucity of research in this area, it is difficult to identify any clear trends regarding the biomechanical factors affected by AGP, however within this review of literature the identified features associated with AGP were most commonly reported at the pelvis and hip. This is perhaps not surprising given the proximity of these segments/joints to the region of pain in AGP.

A number of forms of analysis were reported and investigated in this review. For example, while not examined to date with respect to AGP, there is a growing interest in the association between stiffness and injury. It is possible that if affected with AGP, stiffness may alter the magnitude of loading or the manner in which loads are absorbed by the structures surrounding the pubic symphysis region. The literature review has also highlighted the potential benefits of examining

AGP using a subgroup analysis and the use of continuous waveform approaches in general. Within this thesis, the aforementioned methods will be explored.

Finally, given the association between repetitive loading and chronic overuse injuries such as AGP, there has been a growing interest in the functional role movement variability [both linear and nonlinear (complexity)] may have with respect to injury. The review of literature suggests that variability and complexity are commonly affected by injury and as such the investigation of these features in AGP is warranted.

The primary aim of this thesis was to evaluate the biomechanical factors associated with AGP.

# Chapter 3

# The effects of an exercise intervention on the biomechanics of an athletic groin pain group during a hurdle hop exercise.

## 3.1 Introduction

Athletic groin pain (AGP) is a common chronic injury in sports involving repetitive change of direction, kicking and turning (Thorborg et al. 2017, Orchard, Seward and Orchard 2013, Werner et al. 2009). In male soccer AGP incidence has been reported to account for 4 - 19% of all injuries (Waldén, Hägglund and Ekstrand 2015). Attempts have been made to divide the multiple anatomical presentations of AGP into anatomical regions (Falvey, Franklyn-Miller and McCrory 2009) and entities (Hölmich 2007) in order to improve comparability of papers examining this condition. However, recent work by our group (Falvey et al. 2016, Franklyn-Miller et al. 2017, King et al. 2018), have suggested that this may be arbitrary, as the site of pain is often in exclusion of defined anatomical injury. In this study we

propose to treat AGP as a single entity (much like lower back pain) and do not restrict inclusion by clinical diagnosis.

While the exact anatomical injury underlying AGP remains poorly understood, the modifiable risk factors that are accepted in its development include muscular weakness and imbalance (Thorborg, Branci and Nielsen 2014, Mohammad et al. 2014) reduced muscular activity (Morrissey et al. 2012, Cowan et al. 2004) and reduced range of motion (Nevin and Delahunt 2014, Verrall et al. 2007). As a result of these and other deficits, inappropriate kinematics and kinetics during dynamic agility sporting activities may lead to increased loads and/or unevenly distributed loads across the muscle, tendinous and bony structures of the anterior pelvis, potentially leading to the development of AGP. As such a biomechanical assessment, incorporating an evaluation of change of direction mechanics and joint loading, is an essential step for understanding the biomechanical factors associated with AGP.

Exercise interventions have been shown to be an effective means of rehabilitating AGP, as concluded by recent reviews (Almeida et al. 2013, Jansen et al. 2008, Machotka, Kumar and Perraton 2009, King et al. 2015). To date however, only two studies have examined the effects of an exercise intervention on the biomechanics of an AGP cohort (Gore et al. 2018, [Chapter 4]), (King et al. 2018) and only one of these included an uninjured control group (Gore et al. 2018, [Chapter 4]). This latter study is limited in that it only examined stiffness using a discrete point analysis (Gore et al. 2018, [Chapter 4]). Traditionally, discrete measures (e.g. maximum/minimum values) are chosen for statistical analysis, however this may result in discarding potentially important information contained within the whole waveform (Richter et al. 2014). An alternative to discrete point analysis is to use a continuous analysis approach, which examines the whole biomechanical waveform and avoids assumptions regarding the spatiotemporal foci of which time points to statistically examine.

This study will therefore aim to investigate the kinematic and kinetic variables that change in AGP patients after successful completion of an exercise intervention resulting in pain free return to sport in comparison to a matched uninjured group using a continuous analysis approach. It was hypothesised that the AGP group would be significantly different to the control group pre-rehabilitation in multiple variables, some of these variables would change significantly from pre- to post- rehabilitation and some variables would no longer be significantly different between the AGP group and the controls post-rehabilitation

## 3.2 Method

### 3.2.1 Participants

Data from 65 male subjects with athletic groin pain (mean $\pm$ SD: age 24.6 $\pm$ 4.8yrs, height 180.5 $\pm$ 5.8 cm, mass 81.5 $\pm$ 8.5 kg) who had successfully completed the exercise intervention at the Sports Surgery Clinic, Dublin, Ireland were utilised in this study. This dataset has previously been used solely to examine stiffness (Gore et al. 2018, [Chapter 4]). Within this cohort a primary clinical diagnosis of a pubic aponeurosis injury was made in 46 (71%) cases; hip injury was diagnosed in 14 (22%) cases; inguinal injury was diagnosed in 1 (2%) and adductor injury was diagnosed in 3 (4%) cases. However subject inclusion was not restricted by 'entity' (Franklyn-Miller et al. 2017). Inclusion criteria required all AGP participants to report pain in the anterior hip and groin area during their chosen sporting activity and have symptoms of duration greater than 4 weeks. All AGP subjects were required to have a stated intention of returning to the same level of pre-injury participation in competitive multidirectional sport. AGP subjects were reviewed for clinical history, clinical examination and MRI imaging prior to entry into the study, by a Consultant Physician in Sport and Exercise Medicine as per

(Falvey, King and Kinsella 2015). Subjects with hip joint arthrosis [grade 3 or higher on MRI (Li et al. 1988)], those who did not intend to return to pre-injury activity levels, those who could not commit to completing the rehabilitation programme as prescribed due to time or equipment/facility constraints, and those with underlying medical conditions such as inflammatory arthropathy or infection were excluded. For the control group, 50 male matched uninjured subjects (mean $\pm$ SD: age 23.9 $\pm$ 3.4yrs, height 179.7 $\pm$ 9.26cm, mass 79.8 $\pm$ 13.8 kg) were recruited from local sporting clubs via direct advertisement. Matching criteria were sport, participation level, age and leg dominance. A breakdown of the subject's primary sporting participation is presented in Table 3.1.

Table 3.1: Breakdown of primary sporting participation

| Sporting participation | | | | |
|---|---|---|---|---|
| | AGP | | Uninjured controls | |
| | n | percentage | n | percentage |
| **Gaelic Football** | 46 | 70 % | 28 | 56 % |
| **Hurling** | 7 | 11 % | 4 | 8% |
| **Soccer** | 6 | 10 % | 12 | 23% |
| **Rugby** | 5 | 9 % | 6 | 13% |

The recruitment period was from January 2014 to March 2015. The Sports Surgery Clinic's ethics committee provided ethical approval (REF 25EF011) and all of the participants signed informed consent.

### 3.2.2 Measurements and Protocols

Return to pain free participation in sport, the Copenhagen Hip And Groin Outcome Score (HAGOS) (Thorborg et al. 2011) and adductor squeeze tests (Nevin and Delahunt 2014) were the outcome measures following rehabilitation in this cohort.Patients who demonstrated symptom free completion of the Linear B run-

ning programme and multidirectional drills at maximum intensity were deemed sufficiently rehabilitated to be cleared to return to play.

### 3.2.3 Biomechanical Examination and Data capture

Each subject with AGP attended the biomechanics laboratory on two occasions (pre- and post- intervention) and the control group on one occasion with no intervention. Prior to the experimental testing, the subjects completed a standardized warm-up (Marshall et al. 2015). This involved a three-minute treadmill jog at 8 km/h, five body weight squats and five practice trials of the hurdle hop test.

The lateral hurdle hop involved a lateral hop over a 15 cm hurdle followed by an immediate hop back to the initial starting position (Figure 3.1). The distance between foot contacts was 40 cm. The subject began on their painful leg, contralateral non-weight bearing foot behind with the knee flexed to approximately 90 degrees, and hands unrestricted for balance (Gore et al. 2016). Subjects were instructed to undertake the hop as explosively as possible. The first initial foot contact with the force platform was analysed and three repetitions of the exercise were undertaken to obtain mean scores. A rest period of 30 seconds was taken between repetitions. The subjects had no prior experience of the hurdle hop test.

Reflective markers (14 mm diameter) were placed at bony landmarks on the lower limbs, pelvis and trunk as per the Vicon Plug in Gait model (Vicon Motion Systems, Oxford, UK). Marker position was tracked using 8 infrared cameras (Vicon - Bonita B10, UK), synchronized with two 40x60cm force platforms (AMTI BP400600, USA) collecting ground reaction force data. Motion and force data were captured at a sampling frequency of 200 Hz and 1000 Hz, respectively. Both marker and force data were filtered using a fourth order Butterworth filter with a cut-off frequency of 15 Hz (Kristianslund, Krosshaug and van den Bogert 2013). Kinematic and kinetic calculations were performed in Nexus software

(Vicon Motion Systems, Oxford, UK) and were subsequently exported to Matlab for statistical analysis. Kinematic and kinetic variables were defined as per the standard Vicon Plug in Gait model.

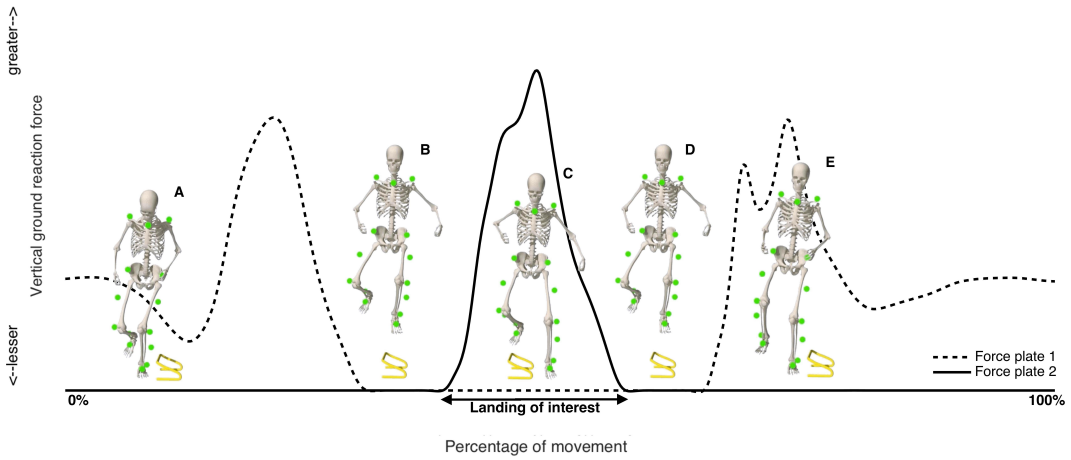


Figure 3.1: Graphical representation of the hurdle hop test. A) Starting position on force plate 1. B) Initial hop over hurdle. C) Initial landing phase that is biomechanically examined on force plate 2. D) Return hop back over the hurdle. E) End position on force plate 1 after hopping back over hurdle.

### 3.2.4 Exercise intervention

All subjects with AGP undertook a rehabilitation programme focused on control of the hip, pelvis and trunk during dynamic loading tasks. The intervention involved three levels of progression (Figure 3.2). Level 1 addressed inter-segmental control and strength, level 2 involved linear running mechanics and increasing linear running load tolerance and level 3 addressed multidirectional mechanics and the transition back to high intensity sprinting. This program was completed 4 days per week alternating between strength and running drills. Advancement through these levels was based on the examining physiotherapist's subjective assessment of the subjects' control, absence of pain during tasks of the preceding level alongside the competence-based assessment of activity. The programme was unsupervised but a physiotherapist assessed each patient's progress at regular intervals (mean $\pm$ SD (Range): 4.92 $\pm$ 1.7 (2-10) patient assessments every 14.29

Figure 3.2: Components of AGP rehabilitation and key performance indicators for progression (King et al. 2018)

$\pm$ 3.9 (9-22) days).Patients who demonstrated symptom free completion of the Linear B running programme and multidirectional drills at maximum intensity were deemed sufficiently rehabilitated to be cleared to return to play. A detailed description of this intervention has previously been published elsewhere (King et al. 2018) in accordance with the TIDieR (template for intervention description and replication) checklist and guide (Hoffmann et al. 2014) and a summary is provided in Appendix C.

### 3.2.5 Data Analysis

Hop height, width and contact time were calculated from force plate data as discrete measurements. Kinematic and kinetic waveforms were examined with Analysis of Characterising Phases (ACP) (Richter, O'Connor, et al., 2014a) [ Ap-

pendix I]. A summary of the biomechanical variables examined in this chapter is presented in Table 3.2.

**Analysis of Characterising Phases** To reduce the dimensionality of the kinematic and kinetic waveform data and identify phases of interest, a VARIMAX rotated principle component analysis was utilised that retains more than 99% of the variance in the data (Richter, O'Connor, et al., 2014a). Once the phases are identified, subject scores were generated for each phase within the magnitude-time domain. These phases were tested using planned t-tests (unpaired t-test: pre-intervention vs. uninjured, post- intervention vs. uninjured; paired t-test: pre- vs. post- intervention). If the phases were significantly different they were retained and extended to either side separately and retested for significance (Richter, O'Connor, et al., 2014a). This was repeated to identify the full phase of the difference, terminating when a data point was not statistically significantly different.

**Statistical Analysis** Both the discrete biomechanical variables and the full phases identified using ACP were tested for significance using planned t-tests (unpaired t-test: pre-intervention vs. uninjured, post- intervention vs. uninjured; paired t-test: pre- vs. post- intervention) and presented along with Cohen's effect size. No multi-comparison adjustment was deemed necessary (Perneger, 1998; Hopkins et al., 2009). Cohen's effect size was reported as small ($< 0.5$), medium (0.5 - 0.8), and large ($> 0.8$) (Cohen, 1988).

Table 3.2: Summary of Biomechanical variables examined

| Variable | Description of Calculation |
|---|---|
| **COM height** | Vertical position of the COM. |
| **COM power ‡** | (Ground reaction force * COM displacement)/time |
| **Ground contact time ‡** | Time from initial contact to toe off defined as when the vertical ground reaction force passes a 5N threshold. |
| **Ankle, Knee and Hip joint angle** | Relevant angle between two adjacent segments described with respect to the proximal coordinate system. |
| **Thorax and Pelvis angle** | Absolute angle relative to global coordinate system. |
| **Joint/ Segment moments** | Calculated using inverse dynamics and presented in the local co-ordinate frame of the distal segment. |
| **Joint/ Segment powers** | Moment * angular velocity |

COM = Centre of mass; ‡ = Manually calculated variable.

## 3.3 Results

Following rehabilitation, all subjects with AGP underwent pain-free return to play (median: 9.14 weeks, range 5.14 - 29.0 weeks), with a significant improvement in maximum adductor squeeze score at 0°, 45° and 90° Table 3.3).

Table 3.3: Maximum pressure achieved during squeeze tests

| Test (mmHg) | Pre | Post | Sig | Cohen's D |
|---|---|---|---|---|
| **Squeeze 90** | 181.09 ± 41.66 | 214.72 ± 36.99 | <0.01 | 0.78 |
| **Squeeze 45** | 221.64 ± 38.81 | 251.04 ± 39.9 | <0.01 | 0.70 |
| **Squeeze 0** | 133.09 ± 29.87 | 141 ± 34.16 | 0.05 | 0.22 |

Squeeze scores measured as (mmHg - millimetres of mercury); D = Cohen's D effect size

HAGOS scores also improved in 5 [Pain ($p < 0.01$, $d = 0.83$), Symptoms ($p < 0.01$, $d = 0.75$), Function in daily living ($p < 0.01$, $d = 0.64$), Function in sport and recreation ($p < 0.01$, $d = 1.13$), Quality of life ($p < 0.01$, $d = 1.01$)] out of the 6 subscales with only one subsection [Participation in Physical Activities ($p = 0.36$, $d = 0.38$)] not changing significantly Table 3.4.

Table 3.4: Results from pre- and post-intervention HAGOS scores

| HAGOS subscale | Pre | Post | Sig | Cohen's D |
|---|---|---|---|---|
| **Pain** | 75.42 ± 12.30 | 87.45 ± 14.53 | < 0.01 | 0.83 |
| **Symptoms** | 64.24 ± 17.29 | 78.27 ± 17.83 | < 0.01 | 0.75 |
| **Function in daily living** | 77.82 ± 15.76 | 88.27 ± 15.23 | < 0.01 | 0.64 |
| **Function in sport and recreation** | 53.54 ± 17.21 | 79.46 ± 21.34 | < 0.01 | 1.13 |
| **Participation in Physical Activities** | 76.58 ± 32.74 | 63.52 ± 36.25 | 0.36 | 0.38 |
| **Quality of Life** | 38.66 ± 15.51 | 60.41 ± 22.84 | < 0.01 | 1.01 |

### 3.3.1 Biomechanical Findings

There were no significant differences in hop height (AGPpre vs. Control p =0.32, d = 0.15, AGPpost vs. Control p = 0.47, d = 0.06, AGPpre vs. AGPpost p =0.43, d = 0.09) or hop width between the groups (pre vs. uninjured: p =0.13, d = 0.11, post vs. uninjured, p = 0.95, d = 0.01, pre vs. post p =0.13, d =0.10).

In total 18 kinematic and kinetic variables were identified in the pre-intervention AGP group that were significantly different to the uninjured group (Table 3.5). Contact time remained statistically slower in the AGP group post-rehabilitation in comparison to the control group. Vertical, medial-lateral and anterior-posterior ground reaction forces remained significantly less post-rehabilitation in the AGP group. There was a significant increase in the AGP centre of mass height from 6-100% of the ground contact phase and post intervention the AGP group were no longer significantly different to the uninjured control group. Vertical centre of mass power absorption from 22-47% and production from 51-74% of the ground contact phase remained significantly lower in the AGP group compared to uninjured values post-rehabilitation. There was a significant reduction in medio-lateral centre of mass power absorption in the AGP group from pre- to post-rehabilitation whilst anterior-posterior centre of mass power absorption remained significantly less in the AGP group post intervention in comparison to the control group.

Pre-intervention, the AGP group had significantly less plantar flexion from 26-76% of the ground contact phase in comparison to the control group, but following the intervention there was no longer a significant difference. In contrast, ankle

plantar flexor moments remained significantly less in the AGP group in comparison to the control group post intervention from 12-92% of the ground contact phase, whilst ankle plantar flexor power absorption and generation also remained significantly less from 8-38% and 47-88% of the ground contact phase, respectively. In the transverse plane there was a trend towards uninjured values with a decrease in ankle external rotator power generation from 31-38% of the ground contact phase. At the knee pre-intervention the AGP group had significantly greater knee extensor moments from 13-24% of the ground contact phase in comparison to the uninjured group and was no longer significantly different post-intervention. In the frontal plane, knee abductor/valgus moments were significantly greater in the AGP group in comparison to the control group pre-intervention from 10-23% and remained so from 7-29% of the ground contact phase post-intervention. Hip abductor moments from 16-22% and hip extensor moments from 44-51% of the ground contact phase were significantly less in the AGP group in comparison to the control group pre-intervention, however post-intervention these phases were no longer significantly different. Pre-intervention hip flexor power absorption from 38-43% was significantly less in the AGP group in comparison to the control group and remained so post-intervention but for a shorter duration. In the transverse plane hip power generation was significantly greater pre-intervention in the AGP group in comparison to the control group from 75-80% of the ground contact phase, but post intervention this was no longer significantly different to the uninjured group. Full kinematic and kinetic waveforms are presented below as mean $\pm$ standard errors ( Figure 3.3 : Figure 3.6).

Table 3.5: Kinematic and Kinetic variables that differed significantly pre-rehabilitation and their changes from pre- to post- rehabilitation.

| | AGPpre vs. Control | | | AGPpre vs. AGPpost | | | AGPpost vs. Control | | |
|---|---|---|---|---|---|---|---|---|---|
| **Variable** | **Percent** | **Sig** | **Cohen's d** | **Percent** | **Sig** | **Cohen's d** | **Percent** | **Sig** | **Cohen's d** |
| **Ground contact time** | - | 0.00 | 0.66 | | | | - | 0.00 | 0.66 |
| **Ground Reaction Force x** | 26-71% | 0.02 | 0.43 | 48-85% | 0.01 | -0.34 | 20-87% | 0.00 | 0.68 |
| **Ground Reaction Force y** | 10-15% | 0.00 | -0.66 | 1-8% | 0.00 | -0.40 | 4-5%<br>10-15%<br>32-37% | 0.03<br>0.00<br>0.04 | 0.41<br>-0.56<br>-0.40 |
| **Ground Reaction Force z** | 23-77% | 0.00 | -0.73 | | | | 24-84% | 0.00 | -0.67 |
| **Centre of mass height** | 31-69% | 0.04 | -0.40 | 6-100% | 0.00 | -0.21 | | | |
| **Centre of mass Power x** | 4-15% | 0.02 | -0.44 | 1-21%<br>43-92% | 0.01<br>0.00 | -0.28<br>0.54 | 56-81% | 0.00 | -0.63 |
| **Centre of Mass Power y** | 10-15% | 0.01 | 0.52 | 88-100% | 0.01 | 0.44 | 1-7%<br>11-16%<br>88-100% | 0.01<br>0.01<br>0.00 | -0.51<br>0.50<br>-0.54 |
| **Centre of Mass Power z** | 21-76% | 0.01 | 0.29 | | | | 22-47%<br>51-74% | 0.00<br>0.02 | 0.04<br>-0.43 |
| **Ankle Angles (Dorsi/Plant)** | 26-76% | 0.02 | 0.46 | | | | | | |
| **Ankle Moment (Dorsi/Plant)** | 12-86% | 0.00 | -0.75 | | | | 13-92% | 0.00 | -0.88 |
| **Ankle Power (Dorsi/Plant)** | 9-33%<br>50-86% | 0.00<br>0.00 | 0.65<br>-0.55 | | | | 8-38%<br>47-88% | 0.00<br>0.00 | 0.73<br>-0.65 |
| **Ankle Power (Int/Ext Rot)** | 31-38% | 0.03 | 0.42 | | | | | | |
| **Knee Moment (Abd/Add)** | 10-23% | 0.03 | 0.42 | | | | 7-29% | 0.01 | 0.49 |
| **Knee Moment (Fle/Ext)** | 13-24% | 0.01 | 0.47 | | | | | | |
| **Hip Moment (Abd/Add)** | 16-22% | 0.01 | -0.47 | 3-9%<br>88-100% | 0.00<br>0.00 | -0.35<br>-0.29 | | | |
| **Hip Moment (Fle/Ext)** | 44-51% | 0.02 | -0.43 | 74-89% | 0.00 | -0.37 | | | |
| **Hip Power (Fle/Ext)** | 38-43% | 0.02 | 0.43 | 13-16%<br>66-81% | 0.04<br>0.01 | 0.39<br>-0.31 | 40-44% | 0.05 | 0.38 |
| **Hip Power (Int/Ext Rot)** | 75-80% | 0.04 | -0.39 | 20-23% | 0.02 | 0.32 | | | |

x = mediolateral, y = anterioposterior, z = vertical

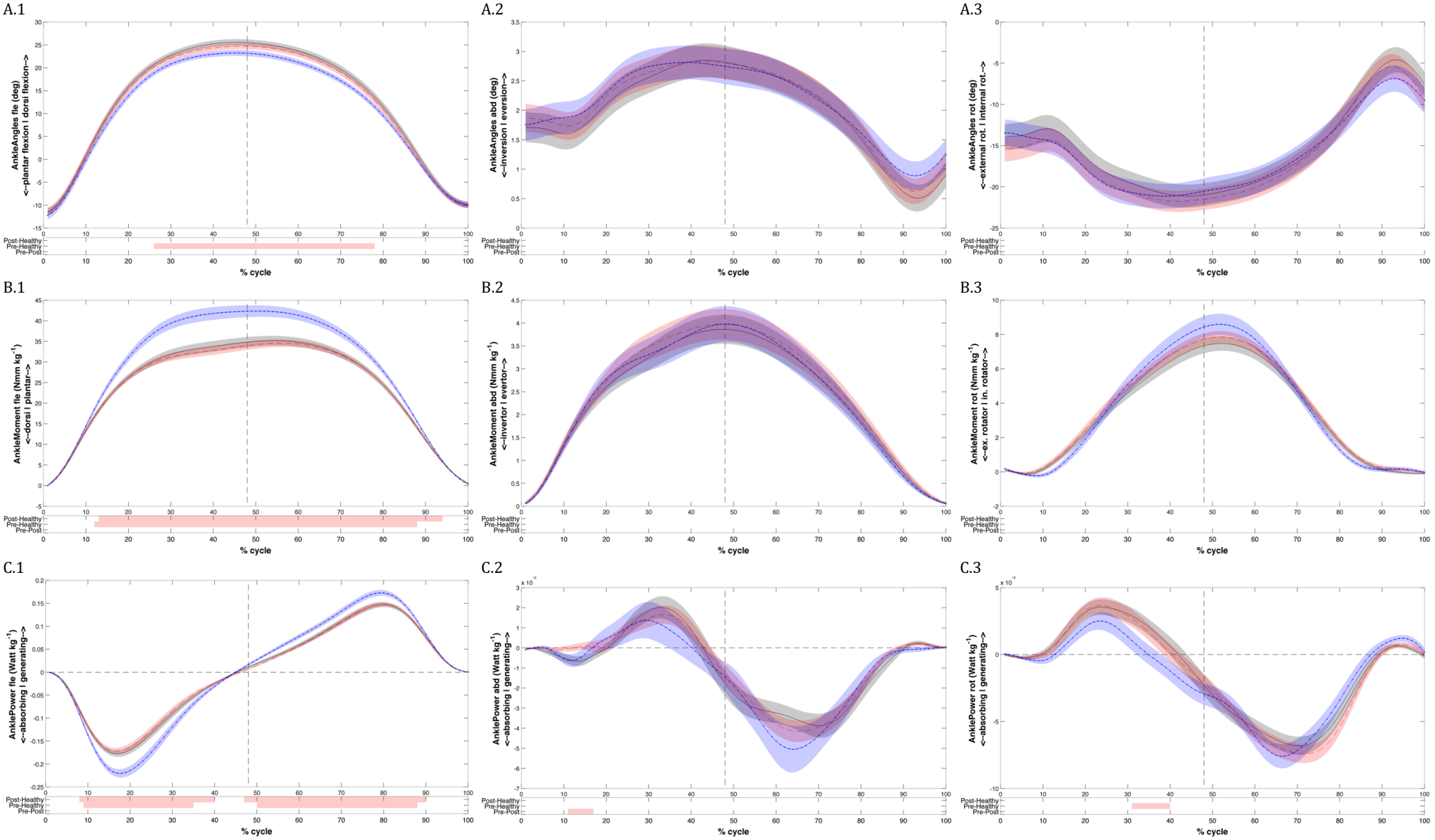


Figure 3.3: Ankle Kinematics and Kinetics. Row A refers to angles, Row B refers to moments, Row C refers to Powers.Column 1,2 & 3 refers to sagittal plane, frontal plane and transverse plane respectively.. (— = Uninjured , — = AGP pre intervention — = AGP post intervention ] Sub graph bar indicates phases of significant difference.

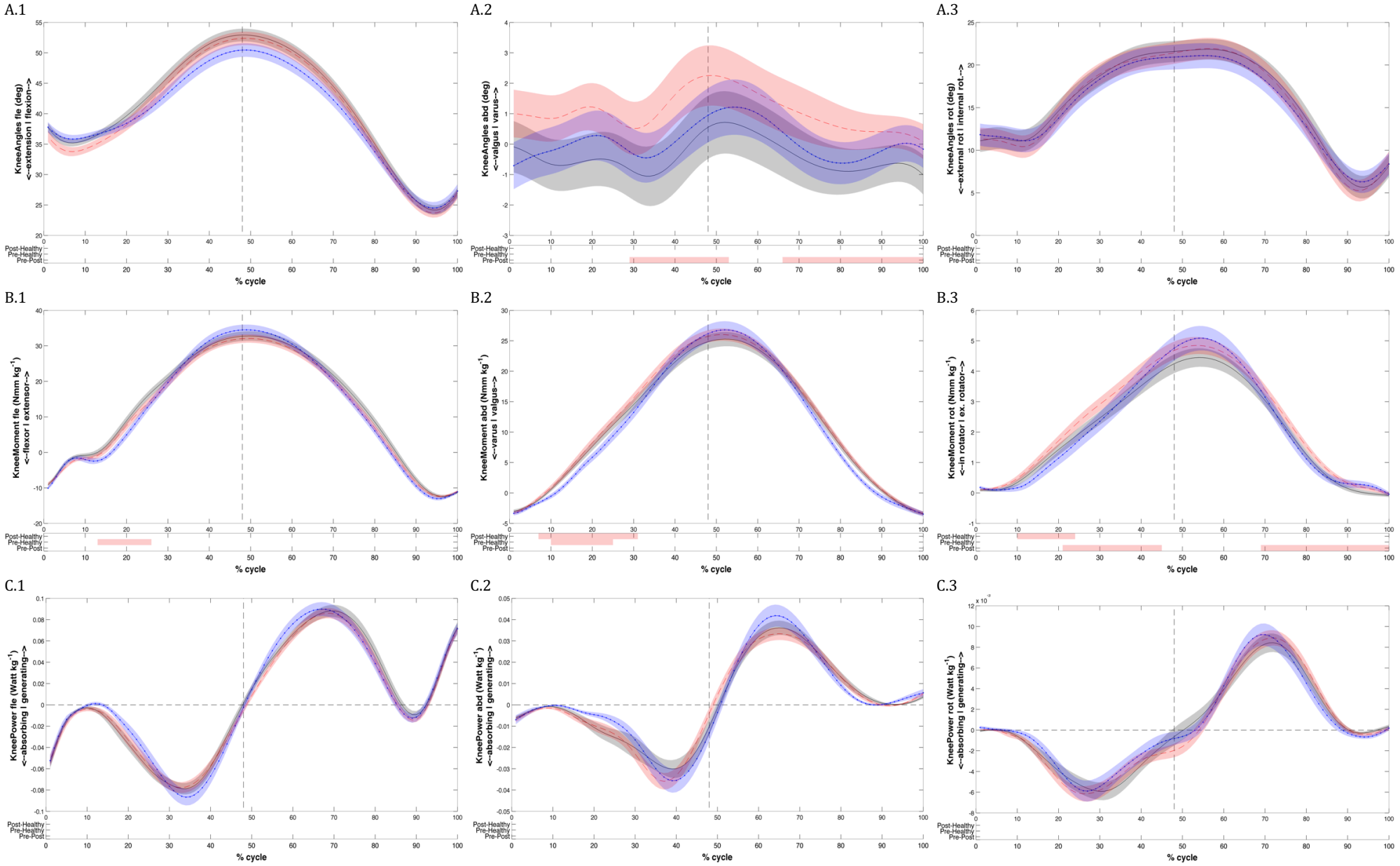


Figure 3.4: Knee Kinematics and Kinetics. Row A refers to angles, Row B refers to moments, Row C refers to Powers.Column 1,2 & 3 refers to sagittal plane, frontal plane and transverse plane respectively.. (— = Uninjured , — = AGP pre intervention — = AGP post intervention ] Sub graph bar indicates phases of significant difference.

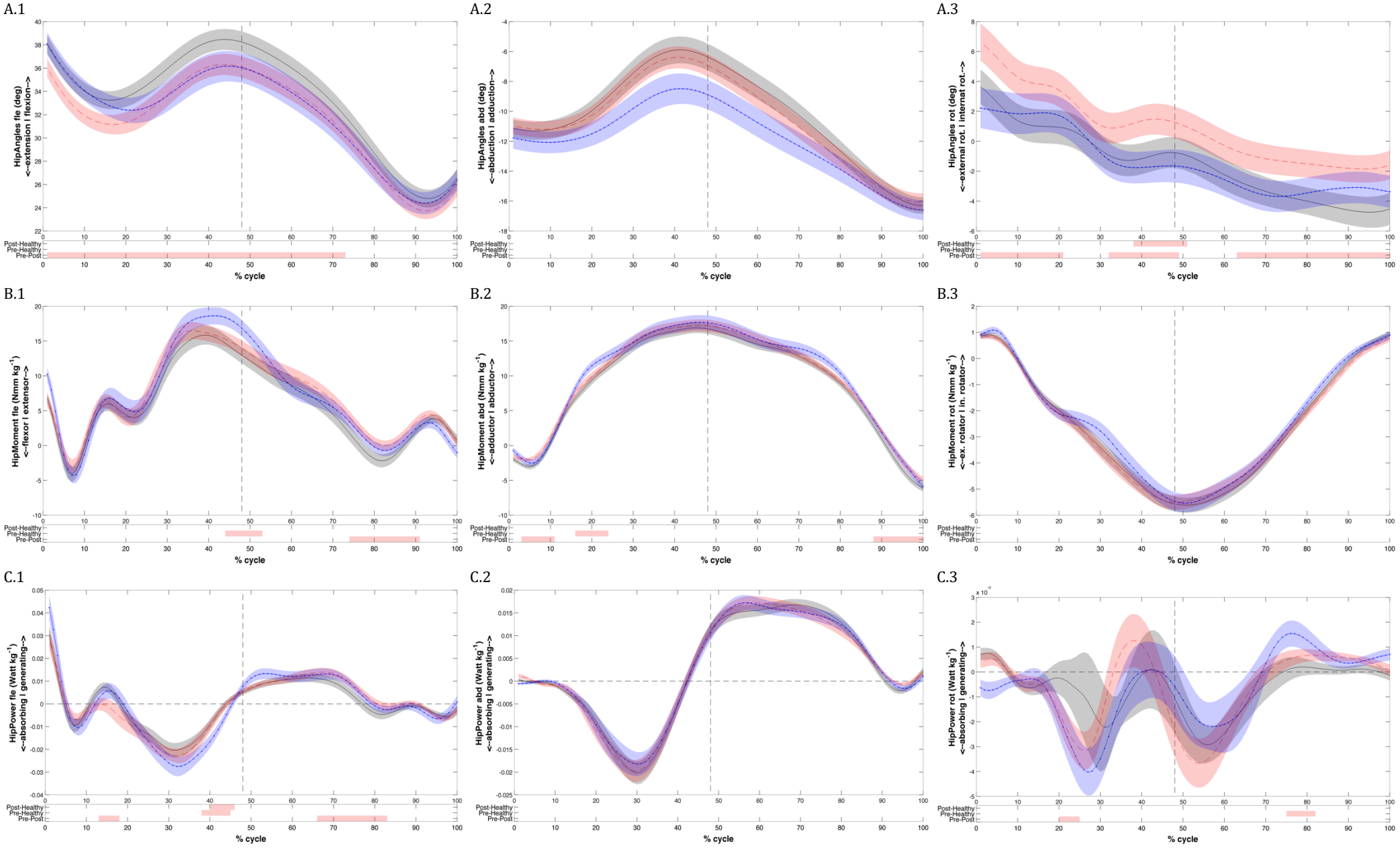


Figure 3.5: Hip Kinematics and Kinetics. Row A refers to angles, Row B refers to moments, Row C refers to Powers.Column 1,2 & 3 refers to sagittal plane, frontal plane and transverse plane respectively.. (— = Uninjured , — = AGP pre intervention — = AGP post intervention ] Sub graph bar indicates phases of significant difference.

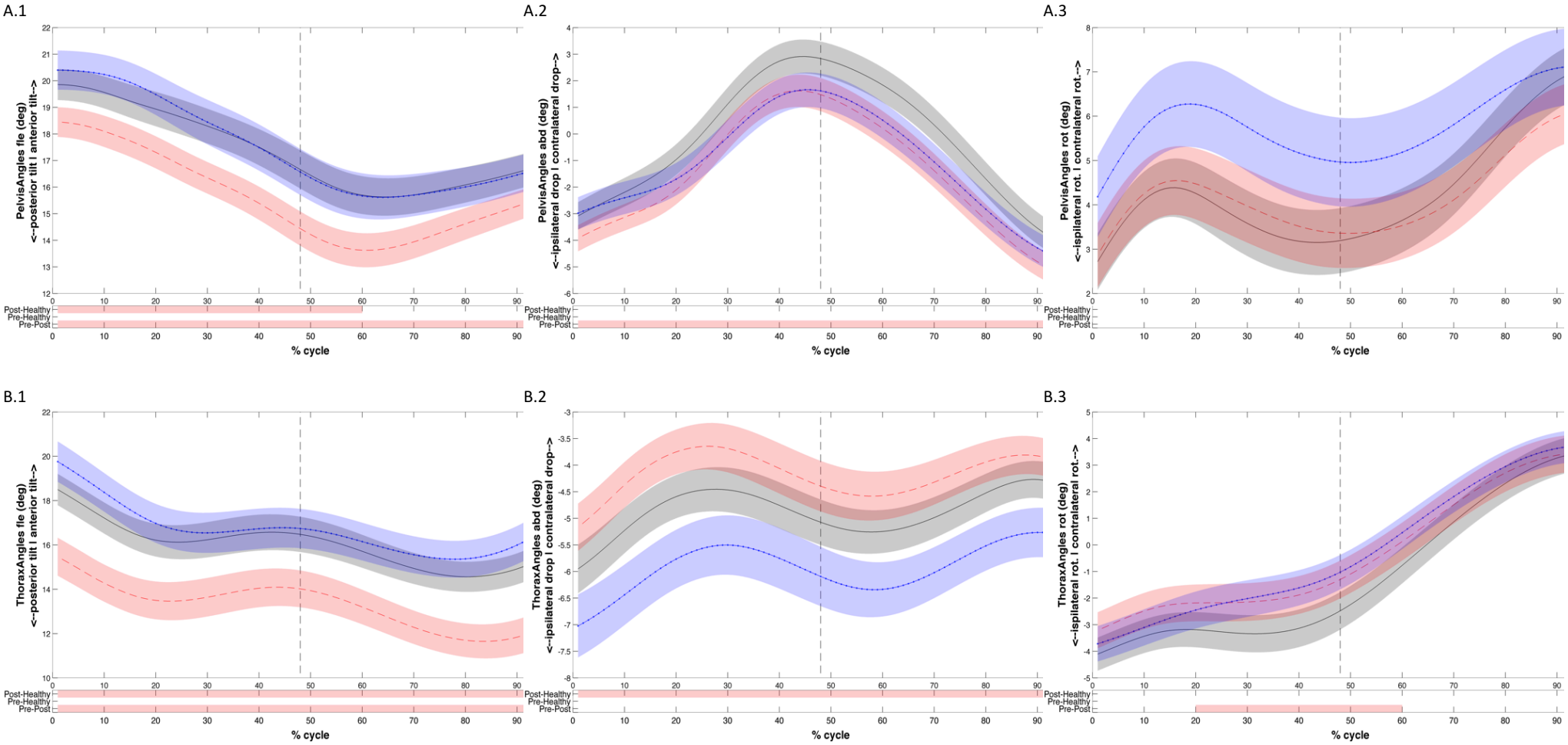


Figure 3.6: Pelvis and Thorax Kinematics.Row A refers to pelvis, row B refers to thorax.Column 1,2 & 3 refers to sagittal plane, frontal plane and transverse plane respectively.. (— = Uninjured , — = AGP pre intervention — = AGP post intervention ] Sub graph bar indicates phases of significant difference.

## 3.4 Discussion

This study presents the effects of an exercise intervention on the biomechanics of an AGP cohort during a hurdle hop task using a continuous waveform approach. While many studies utilise a return to sport alone as a measure of successful rehabilitation, this study also adopted adductor squeeze tests (Nevin and Delahunt, 2014) and a patient reported outcome measure (HAGOS) as outcome measures (Thorborg et al., 2011). All AGP patients within this cohort were returned to play in a median time of 9.14 weeks with no return of symptoms. A total of 18 kinematic and kinetic variables were identified which were significantly different between the subjects with AGP pre-intervention and the control group. These identified variables were then subsequently examined to assess for change post-intervention. As a result of the intervention, eight of these variables were no longer significantly different between the AGP group post-intervention and the controls. It would seem logical therefore to suggest that these eight variables were the most related to return to play in this cohort.

**Biomechanical Findings** Before rehabilitation hip abductor moments were significantly less in comparison to the uninjured controls. As noted by our research group (Gore et al. 2018, [Chapter 4]) this may be utilized as a compensatory mechanism in order to reduce ipsilateral hip joint reaction forces (Neumann, 2010; Wesseling et al., 2015) or may be a risk factor due to increased loading at a localised soft tissue level. While the observed reduction in hip abductor moments may be explained by increased hip adductor moment, previous research has also demonstrated reduced gluteus medius activity in AGP patients (Morrissey et al., 2012). Post-rehabilitation there was a restoration of hip abductor moments and the AGP group was no longer significantly different to uninjured controls. This is in line with work by King et al. (2018) who reported reduced adductor moments

in AGP patients post-rehabilitation during a running cut task. Furthermore, at the hip following rehabilitation there was an increase in hip external rotation power production and hip extensor moments, with the AGP group no longer significantly different from the uninjured controls. Contrary to previous research (Kopper, Ureczky and Tihanyi, 2012), this increase in hip extensor moments was observed even with a significant decrease in thorax flexion angle from pre- to post-rehabilitation suggesting that the hip extensor muscles are working more post-intervention. Despite this, hip extensor power dissipation remained significantly less post-rehabilitation alluding to a reduced joint angular velocity and potentially a protective mechanism utilized to reduce loading at the hip/pelvic region.

At the knee, there were significantly greater eccentric moments in the AGP group pre-rehabilitation in comparison to the uninjured group. Post-intervention however, knee moments were no longer significantly different suggesting a shift to a hip dominant strategy.

Similar to previous research (Gore et al. 2018, [Chapter 4]), this present study observed some of the greatest effect sizes between the AGP group pre-rehabilitation and the uninjured controls at the ankle joint. Despite an effective rehabilitation, the ankle sagittal plane moments and powers remained significantly less in the AGP group in comparison to the uninjured group post-rehabilitation. The ankle joint is often over-looked in injuries involving proximal joints, however previous research has indicated that the ankle joint contributes 45.7% of total energy dissipation during single leg landings (Yeow, Lee and Goh, 2011). Furthermore, the ankle joint may be a major modulator of lower limb stiffness (Farley and Morgenroth, 1999). Whilst increased stiffness has been positively related to increased peak force in landing and thus a risk for injury (Devita and Skelly, 1992), too little stiffness may also lead to injury with excessive joint motion leading to soft tissue injury (Williams, McClay and Hamill, 2001; Granata, Padua and

Wilson, 2002). Future research should attempt to ascertain if the observed altered ankle function in the present study is an unaddressed risk factor for AGP, or a protective mechanism. Regardless, it is possible that the efficacy of the reported rehabilitation intervention could be further enhanced with a focus on improving ankle function.

Following rehabilitation, center of mass height from 31-69% of the ground contact phase was no longer significantly less compared to the control group, due to the significant reductions in trunk flexion, ankle dorsiflexion and hip flexion angles in the AGP group from pre- to post-rehabilitation. Despite there being no significant difference in hop height or hop width between the AGP group pre-rehabilitation and the controls, the magnitude of the ground reaction force was significantly less in the AGP pre-rehabilitation in comparison to the uninjured controls. This may be explained by the impulse momentum relationship, whereby the AGP achieved the same hurdle hop task by increasing the ground contact time over which the force is applied. Ground reaction force in all three planes, and centre of mass power in the vertical and medio/lateral direction remained significantly less in the AGP group post-rehabilitation in comparison to the uninjured control group, while contact time remained significantly longer in the AGP group post-rehabilitation.The biomechanical factors that were significantly different between those with and with AGP pre- rehabilitation, but remained unchanged with rehabilitation, may represent either unaddressed risk factors or compensatory mechanisms for AGP. Prospective research is required to conclusively determine which of the above biomechanical measures are risk factors for AGP and should be targeted in rehabilitation.

**Clinical Findings** The findings from this study demonstrate that exercise therapy can be an effective means of treatment for AGP, as supported by previous research (Holmich et al., 1999; Weir et al., 2011; King et al., 2018). The adductor

squeeze scores statistically improved at 0°, 45° and 90° in the AGP group from pre- to post-rehabilitation and the HAGOS questionnaire statistically improved in five out of the six sub scores. While previous research has demonstrated that with an effective AGP rehabilitation HAGOS scores can remain reduced compared to athletes who have never had groin pain (Thorborg, Rathleff and Petersen, 2015; King et al., 2018), it is uncertain why the Participation in Physical Activities subsection of the HAGOS questionnaire remained unchanged with rehabilitation in this study. It is possible that this may reflect patients returning to full participation in sport cautiously, and only being cleared to return to play at the time of collecting post-intervention HAGOS scores. The return to play time of 9.14 weeks reported in this present study is similar to that of (King et al., 2018) [9.9 weeks], which implemented the same rehabilitation protocol in another cohort of AGP patients. Interestingly, the return to play time reported in this study is approximately half that of the times reported by (Holmich et al., 1999) [18.5 weeks] and (Weir et al., 2011) [17.3 weeks]. Whilst it should be noted that both of the aforementioned studies are randomised control trials, it is plausible that quicker return to play time reported in both this present study and that of (King et al., 2018) was due to the rehabilitation being more targeted in nature with a focus on whole body movement. Indeed AGP case series studies involving a high degree of individualisation, as employed in the present study, reported generally improved return to play times (4 - 14 weeks) (Rodriguez et al., 2001; Wollin and Lovell, 2006; Jarosz, 2011; Vijayakumar, Nagarajan and Ramli, 2011; Jard et al., 2014; McAleer et al., 2015). Unlike these case series studies, the current study's intervention was unsupervised in nature. Previous research has demonstrated that supervised interventions may be superior in outcome compared to unsupervised home-based rehabilitation (Feger et al., 2015). As such it is possible that return to play times may have differed further under a directly supervised rehabilitation program.

### 3.4.1 Limitations and future research

Given the focus of this paper on identifying biomechanical factors associated with athletic groin pain, only patients who completed the rehabilitation protocol and returned for post intervention testing were included in this study. Previous research has demonstrated a 73% return to play rate using the same exercise intervention presented in this study (King et al., 2018). As such the clinical outcome (100% of patients return to sport) is biased, and important information regarding the mechanics of those who do not successfully rehabilitate is not gained from this study. Rather than a time-based approach to rehabilitation progression, the intervention presented within this current study was dependent on achieving set criteria. Whilst this is a more ecologically valid approach, it is important to note that the variation in reassessment time post-rehabilitation may be a confounding factor when examining how change in biomechanics is associated with return to play in this cohort. Although outside the scope of this paper, given the large standard deviations present in the data, it is possible that the presence of sub-groups demonstrating distinct movement patterns may have masked some changes produced by the intervention. Our research group have recently identified the presence of sub-groups in AGP patients during a running cut task (Franklyn-Miller et al., 2016). Further research should consider replicating this current study in conjunction with a cluster analysis. Finally, it has been suggested that segmental co-ordination and its variability may be more discriminative between normal and pathological movement than single joint analysis as conducted in this study (Hamill, Palmer and Van Emmerik, 2012). Future research should explore segmental co-ordination variability in AGP patients to determine if it is of importance in the examination of AGP.

## 3.5 Conclusion

The AGP patients in this study were returned to play in a median time of 9.14 weeks with no return of symptoms. This successful return to play was associated with significant improvements in five out of six HAGOS subsections and a significant improvement in adductor squeeze scores. A total of 18 kinematic and kinetic variables were identified which were significantly different between the subjects with AGP pre-rehabilitation and the control group. Following rehabilitation eight of these variables [centre of mass height (31-69% of the ground contact phase), centre of mass frontal plane power (4-15% of the ground contact phase), ankle flexion angle (26-76% of the ground contact phase), ankle rotation power (31-38% of the ground contact phase), knee extensor moments (13-24% of the ground contact phase), hip abduction moment (16-22% of the ground contact phase), hip extensor moment (44-51% of the ground contact phase) and hip rotation power (75-80% of the ground contact phase)] were no longer significantly different between the two groups. These variables may represent the factors most related to return to play in this cohort and are potential targets for rehabilitation. Clearly however, this needs to be re-examined using appropriate prospective research to determine more conclusively what biomechanical factors are related to the development of this condition.

### 3.5.1 Link between Chapter 3 and Chapter 4

Chapter 3 sought to investigate the biomechanics of AGP using a continuous waveform approach. As a result of this study, eight variables were identified which may represent targets for AGP rehabilitation.

Recently however, there has been a growing interest in the role stiffness may have to play with respect to the pathomechanics of AGP. Unlike the more traditional approach taken in Chapter 3 which examines moments and angles in

isolation, Chapter 4 utilises summative measures of stiffness, calculated as the ratio of joint moment to angular joint displacement and represents resistance of a body to deformation under a given load (Pearson and McMahon, 2012). Stiffness may be of importance to the development of AGP, since any alteration in the magnitude of loading or the manner in which loads are absorbed may overload the musculotendinous and bony structures surrounding the pubic symphysis region (Meyers, Greenleaf and Saad, 2005; Franklyn-Miller et al., 2017). Chapter 4 will be explored utilising the same dataset used in Chapter 3.

# Chapter 4

# Is stiffness related to athletic groin pain?

This study has been published within the Scandinavian Journal of Medicine & Science in Sports:

Gore, S.J., Franklyn-Miller, A., Richter, C., Falvey, E.C. King, E., and Moran, K.A., 2018. Is stiffness related to athletic groin pain ? Scandinavian Journal of Medicine & Science in Sports,In press.

It is presented here in this chapter with only minor changes to confirm to the style and formatting of this thesis.

## 4.1 Introduction

Athletic groin pain (AGP) is prevalent in field sports with recurrent accelerations, decelerations and changes of direction (Ekstrand and Hilding, 1999; O'Connor, 2004; Werner et al., 2009; Murphy et al., 2012; Blake et al., 2014). Despite this, the biomechanics contributing to AGP remains poorly understood and under

investigated in comparison to other sporting injuries, such as anterior cruciate ligament injury (Alentorn-Geli et al., 2009) and patella femoral pain syndrome (Neal et al., 2016).

Stiffness, which is resistance of a body to deformation under a given load (Pearson and McMahon, 2012), has attracted attention in injury prevention research (Milner et al., 2006; Hamill, Moses and Seay, 2009; Serpell et al., 2012) as a potentially modifiable risk factor (Brazier, Bishop and Simons, 2014).

Two types of stiffness are typically measured when examining dynamic athletic tasks, whole body vertical stiffness and joint stiffness. Whole body vertical stiffness reflects the resistance of the centre of mass to vertical displacement under a given vertical ground reaction force (McMahon and Cheng, 1990). It is often utilised to represent the stiffness of the lower extremity as a whole (McMahon and Cheng, 1990; Farley, Glasheen and McMahon, 1993; Maquirriain, 2012; Brauner et al., 2014). In contrast, joint stiffness refers to the resistance of a particular joint to rotation under a given moment of force (Farley et al., 1998). The majority of studies examine whole body stiffness, whereas an examination of joint stiffness is advantageous in exploring the contribution of each joint to the sum of whole body stiffness. It is important to note that stiffness examined at joint level is not true stiffness but rather is know as a 'quasi-stiffness'. The distinction between quasi-stiffness and stiffness was discussed in length by Latash and Zatsiorsky (1993) and outlined in both section 2.4 and Appendix E of this thesis, however for simplicity the term 'stiffness' is utilised throughtout this chapter.

Stiffness can be modulated by the central nervous system to maintain the dynamics of locomotion in response to changes in the environment (Farley et al., 1998) and task demands (Farley and Gonzlez, 1996; Brughelli and Cronin, 2008). Indeed, it has been demonstrated that stiffness in individual joints of the lower extremity may change when running under varying conditions while whole body stiffness can remain constant (Hamill, Gruber and Derrick, 2014). As such, it is

likely that joint stiffness reflects more closely, localised regions of loading than whole body stiffness. Further, it has been suggested that abnormal magnitudes of stiffness may lead to an increased risk of injury by increasing peak force and/or rate of force development (Milner et al., 2006), or conversely by increasing the energy absorbed by soft tissues in a lengthened position (Butler, Crowell and Davis, 2003). In line with this, authors have suggested that high levels of stiffness may be related to bony injuries, while too little stiffness may result in soft tissue injury (Williams, McClay and Hamill, 2001; Butler, Crowell and Davis, 2003; Williams et al., 2004). Previous research has also associated greater stiffness with stress fracture (Milner, Hamill and Davis, 2007), and contrastingly lesser stiffness with Achilles tendinopathy (Maquirriain, 2012). With respect to AGP, stiffness may be of particular importance as any alteration in the magnitude of loading or the manner in which loads are absorbed may overload the musculotendinous and bony structures surrounding the pubic symphysis region (Meyers, Greenleaf and Saad, 2005; Franklyn-Miller et al., 2017). To date however, no research has examined whether stiffness is affected by AGP.

In light of the challenges associated with completing prospective research, identifying factors truly associated with an injury is not often possible. As such, the biomechanical comparison of injured versus uninjured participants is a common research approach (Drewes et al., 2009; Morrissey et al., 2012; Nevin and Delahunt, 2014). While this case-control approach is useful, the findings may not be deterministic of injury (e.g. differences may be an outcome of the injury itself), thereby limiting the application of their findings. An alternative, but much less common approach is to examine the biomechanical changes that are associated with an effective rehabilitation programme (as determined by achieving return to play status). In line with the probabilistic approach to causation (Spirtes, Glymour and Scheines, 2000; Marshall and Moran, 2015), the biomechanical factors that change with a successful rehabilitation are possibly causative of the

improvements in injury status and therefore more likely associated with the injury. This approach has been utilised with respect to both identifying targets for rehabilitation (Snyder et al., 2009; Thorp et al., 2010; Ferber, Kendall and Farr, 2011; King et al., 2018) and training-interventions for performance (Sheppard et al., 2009; Marshall and Moran, 2015). Examining the changes that occur with rehabilitation is not without its own limitations of course (e.g. it is possible that the observed biomechanical changes following rehabilitation are not related to the underlying injury, but simply associated with the non-rehabilitative effect of the exercise). A more robust approach would therefore be to combine the case-control analysis (injured versus uninjured), the pre- versus post-rehabilitation analysis and a post-rehabilitation versus uninjured analysis. Logically, the biomechanical factors identified in the case-control analysis (pre- rehabilitation vs. uninjured), that change with successful rehabilitation (pre- vs. post- rehabilitation) to become more similar to the uninjured group (post- rehabilitation vs. uninjured), are more likely to be related to the underlying injury than any of the above approaches in isolation. To date however, this form of analysis has not been conducted in AGP research.

When investigating stiffness, sagittal plane actions, such as running (Coleman, Cannavan and Horne, 2012) hopping (Chelly and Denis, 2001) and drop jump tasks (Maloney, Fletcher and Richards, 2015) are the most commonly used activities examined in this area. However, field sports require dynamic actions not confined to the sagittal plane (Bloomfield, Polman and O'Donoghue, 2007), and these multi-planar movements are common in sports where AGP is a common presentation (Ekstrand and Hilding, 1999; O'Connor, 2004; Werner et al., 2009; Murphy et al., 2012; Blake et al., 2014). A movement stressing frontal and sagittal plane control such as a lateral hurdle hop task (Marshall et al., 2015; Gore et al., 2016) may thus be a more effective screening test to examine stiffness qualities in AGP patients.

The aim of this study was to determine if AGP affects whole body vertical and/or joint stiffness and if so whether return to play following rehabilitation is associated with a change in stiffness.

It was hypothesized that (a) prior to rehabilitation the AGP group would be less stiff in comparison to the control group, and (b) that stiffness would increase from pre- to post-rehabilitation intervention to become more similar to the uninjured group.

## 4.2 Methods

### 4.2.1 Participants

Sixty-five male subjects with AGP who had successfully completed the exercise intervention at the Sports Surgery Clinic, Dublin, Ireland were examined in this study, along with 50 male matched uninjured controls that were recruited from local sporting clubs by direct advertisement. The recruitment period was from January 2014 to March 2015 and comprised of the retrospective inclusion of subjects with AGP who had successfully completed the exercise intervention during this time frame. Inclusion criteria required all AGP participants to undergo clinical consultation, MRI imaging and physical examination to confirm diagnosis of AGP as per criteria previously published (Falvey, King and Kinsella, 2015). Additional inclusion criteria required all participants to be between the ages of 18 $\pm$ 35 and involved in multidirectional field sports. In light of recent research by our group (Franklyn-Miller et al., 2017) we treat AGP as a single entity (much like lower back pain) and do not restrict inclusion by 'entity'.

Exclusion criteria for AGP participants included the presence of hip joint arthrosis, an underlying medical condition such as inflammatory arthropathy or infection, symptoms less than four weeks, lack of intent to return to pre-

injury activity levels, and those who did not successfully complete the exercise rehabilitation program. The control group were uninjured but matched to the AGP group based on age, sport, and participation level alongside leg dominance. The Sports Surgery Clinic ethics committee approved the study (REF 25EF011) and all of the participants signed informed consent.

### 4.2.2 Measurements and Rehabilitation Protocol

AGP subjects completed a three stage rehabilitation programme focusing on inter-segmental control and strength, linear running mechanics and change of direction mechanics, as previously published (King et al., 2018), and detailed in appendix (C). Components of strength, power and plyometric training were incorporated into the rehabilitation program, which as noted in a recent review are all effective means of increasing lower limb stiffness (Brazier, Bishop and Simons, 2014). No aspect of the rehabilitation program was specifically targeted according to pre-rehabilitation stiffness values, but was applied generally to the cohort. The programme was unsupervised but a physiotherapist assessed each patient's progress at regular intervals [mean $\pm$ SD (range): 4.92 $\pm$ 1.7 (2 - 10) patient assessments every 14.29 $\pm$ 3.9 (9 - 22) days]. Progression from level 1 to level 2 of the rehabilitation program was indicated once the patients achieved a negative crossover sign as determined by a lack of pain in the contralateral limb during resisted hip flexion (Brukner and Khan, 2012). Patients progressed from level 2 to level 3 once the subject achieved symmetrical internal hip rotation at 90°, pain-free squeeze at 45° and symptom free completion of the Linear A running programme. Patients who demonstrated symptom free completion of the Linear B running programme and multidirectional drills at maximum intensity were deemed sufficiently rehabilitated to be cleared to return to play. No follow up post return to play was conducted within this cohort. The Copenhagen Hip And

Groin Outcome Score (HAGOS) (Thorborg et al., 2011) were examined pre- and post-rehabilitation.

### 4.2.3 Biomechanical Examination

Prior to the experimental testing the subjects completed a standardised dynamic warm-up (Marshall et al., 2014) involving a three-minute treadmill jog at 8 km/h, five body weight squats and five practice trials of the hurdle hop test. The hurdle hop involved a lateral hop over a 15 cm hurdle followed by an immediate hop back to the initial starting position (Figure 4.1). The distance between foot contacts was 40 cm (the distance between force plate centres). Reliability for the lateral hurdle hop had previously displayed good to excellent reliability (Cicchetti, 1994) with a median reliability coefficient of 0.89 (range 0.67 to 0.97) for the biomechanical measures which contribute to the calculation of stiffness (moments and angles) (Marshall et al., 2015). During testing, the subjects wore their own athletic shoes, and were instructed to wear the same footwear for both testing sessions where applicable. The AGP participants were examined on their painful side, contralateral non-weight bearing foot behind with the knee flexed to approximately 90 degrees, and hands unrestricted for balance. Of the 65 AGP patients 17% had bilateral groin pain. Where the AGP patient had bilateral pain, the leg examined was chosen at random as previously described (Franklyn-Miller et al., 2017). Three repetitions of this test were undertaken to obtain mean scores. Participants were instructed to undertake the hop as quickly as possible and it was the first initial landing phase that was analysed (Figure 4.1). Subjects with AGP were tested pre-(AGP pre) and post-rehabilitation (AGP post), while controls were tested once. The post-rehabilitation test was conducted on the day of completion of the rehabilitation protocol.

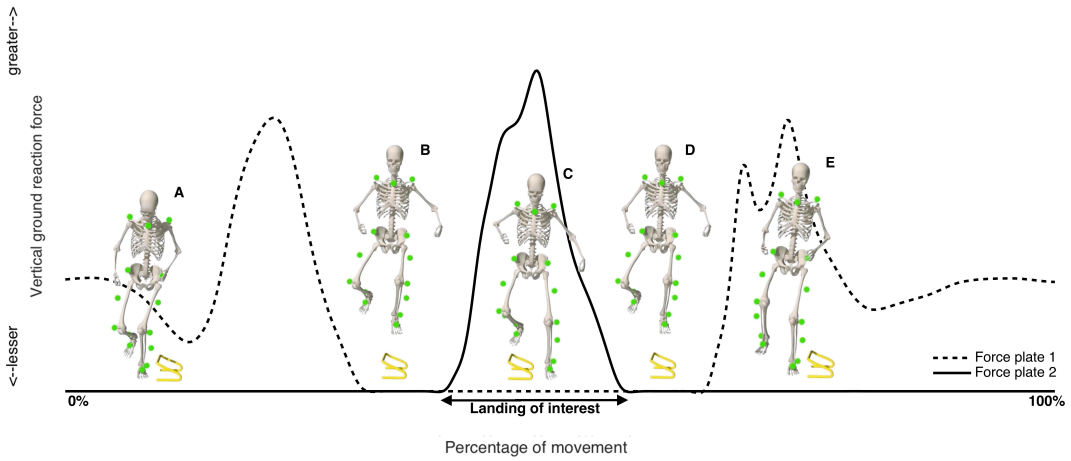


Figure 4.1: Graphical representation of the hurdle hop test. A) Starting position on force plate 1. B) Initial hop over hurdle. C) Initial landing phase that is biomechanically examined on force plate 2. D) Return hop back over the hurdle. E) End position on force plate 1 after hopping back over hurdle.

### 4.2.4 Data Capture

Reflective markers (14 mm diameter) were placed at bony landmarks on the lower limbs, pelvis and trunk as per the Vicon Plug in Gait model (Vicon Motion Systems, Oxford, UK) (Marshall et al., 2014). Lower extremity kinematics and kinetics were captured. Three-dimensional marker position was tracked using 8 infrared cameras (Vicon - Bonita B10, UK), synchronized with two 40x60 cm force platforms (AMTI - BP400600, USA) collecting ground reaction force data. Motion and force data were captured at a sampling frequency of 200 Hz and 1000 Hz, respectively.

### 4.2.5 Data Processing

Both marker and force data were filtered using a fourth order Butterworth filter with a cut-off frequency of 15 Hz (Kristianslund, Krosshaug and van den Bogert, 2013). Moment and angle calculations were performed in Nexus software (Vicon Motion Systems, Oxford, UK). The data was subsequently exported to Matlab 2013b (Mathworks, USA) where stiffness was calculated and the statistical analysis conducted. Stiffness was examined during the eccentric phase of the hurdle hop

action defined as the period from initial ground contact to peak whole body negative power. Negative whole body power was calculated as the external work done per unit time in an attempt to bring the body to a resting state. Whilst many studies have calculated stiffness from initial contact to peak vertical ground reaction force (Brauner et al., 2014; Maloney, Fletcher and Richards, 2015), the authors felt the eccentric phase more useful given that the timing of peak vertical ground reaction force varied considerably (Mean ± SD: 41.3 ± 12.0% of total ground contact), with 33% of all participants producing their peak force during the whole body concentric phase. When examining the moment-angle waveforms of the hurdle hop task in this present study, not all joints demonstrated a clear biphasic pattern as presented in previous research (Farley and Morgenroth, 1999; Kuitunen, Komi and Kyrölinen, 2002; Hamill, Gruber and Derrick, 2014). Indeed many moment-angle waveforms demonstrated a polyphasic pattern (Figure 4.2) with fluctuating moment signs (e.g. between positive and negative abductor moments).

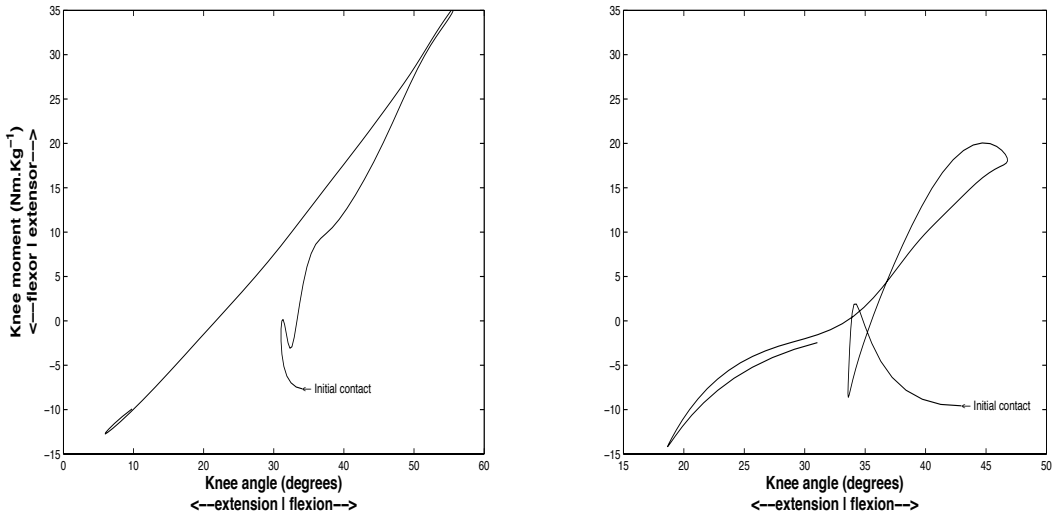


Figure 4.2: Graph depicting a typical biphasic moment-angle pattern as presented in the literature (left) and a polyphasic pattern often identified in this present research (right).

In light of these fluctuations, measuring a mean stiffness across varying signs of net moments would produce an erroneous measure of stiffness containing joint

moments that are not functionally comparable. For this reason, the current study examined joint stiffness from initial contact to peak whole body negative power, for phases when the most prevalent eccentric moments were acting (Figure 4.3). To allow comparison across subjects, the most prevalent eccentric moment was identified at a group level and a moment was deemed to be acting eccentrically when the net joint moment was acting in opposition to the angular displacement of the joint. In the sagittal plane, this involved calculating stiffness for every phase where the extensor/plantar-flexor moments acted eccentrically. In the frontal plane, eccentric hip, knee and ankle abductor/evertor moments were examined. In the transverse plane stiffness was only calculated when the internal rotator moments acted eccentrically at the hip and ankle while at the knee, eccentric external rotator moments were examined. For example, in Figure 4.3 for the calculation of hip extensor stiffness the previously defined conditions were met at two phases.

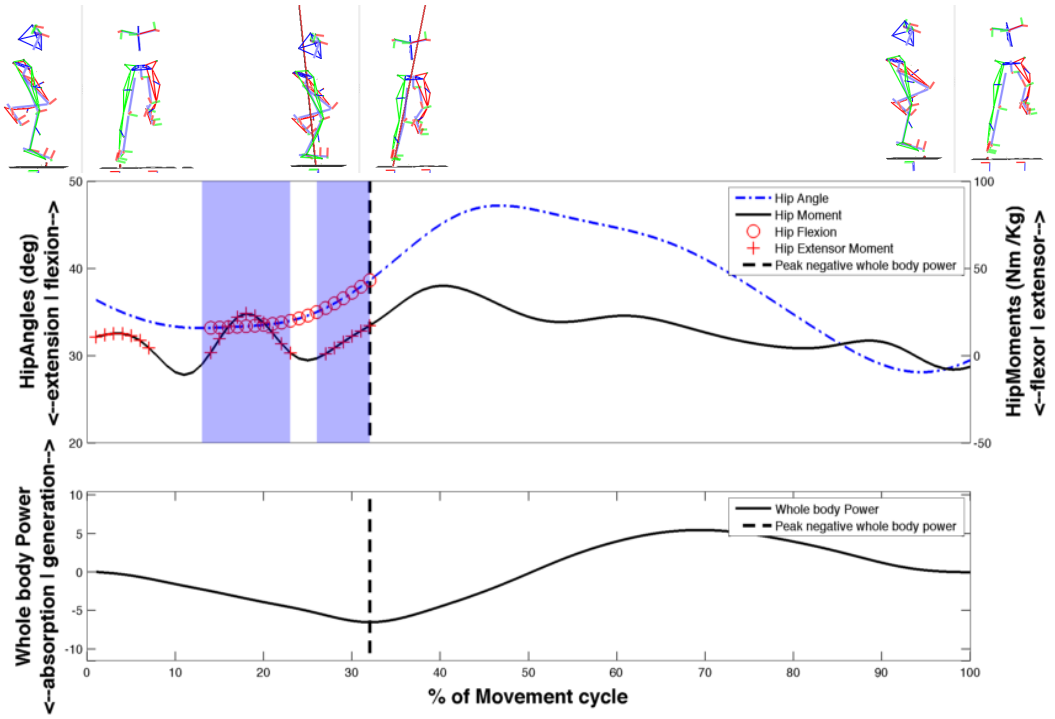


Figure 4.3: Hip flexion/extension angles (°) and Hip flexor/extensor moments (Nm/kg) plotted for a single trial from initial contact to toe off. Blue shaded region indicates the phases that adhere to the condition of an extensor

For every phase i where these conditions were met, joint stiffness at the hip, knee and ankle was calculated as the ratio of the change in joint moment to the change in joint angle for all three planes, from the first to last data point within phase i. Similarly, vertical stiffness was calculated as the ratio between the change in vertical ground reaction force and the vertical displacement of the centre of mass. Both joint and vertical stiffness were subsequently presented normalised to body mass (Equation 4.1 and Equation 4.2).

$$Normalised\ whole\ body\ vertical\ stiffness = \frac{\Delta F_z.Kg^{-1}}{\Delta COM_z} \tag{4.1}$$

$$Normalised\ joint\ stiffness\ (NJS_i) = \frac{\Delta M.Kg^{-1}}{\Delta \theta} \tag{4.2}$$

Where $\Delta F_z$ is the change in vertical ground reaction force, $\Delta COM_z$ is the displacement of the centre of mass, $\Delta M$ is the change in joint moment and $\Delta \theta$ is the range of motion of the joint. To adequately represent the mean stiffness of a joint, a weighted mean was required to account for the intermittent nature and varying durations of the included phases (Equation 4.3):

$$Mean\ weighted\ net\ joint\ stiffness\ (NJS_i) = \frac{\sum_{i=1}^{i_{max}} NJS_i * \Delta\theta}{\sum_{i=1}^{i_{max}} \Delta\theta} \tag{4.3}$$

Hop height and hop width were calculated as the vertical and horizontal distance the centre of mass travelled from the point of take-off to the time the centre of mass reached its highest point. A summary of the biomechanical variables examined in this chapter is presented in Table 4.1.

Table 4.1: Summary of Biomechanical variables examined

| Variable | Description of Calculation |
|---|---|
| **Ankle, Knee and Hip joint angle** | Relevant angle between two adjacent segments described with respect to the proximal coordinate system. |
| **Joint moments** | Calculated using inverse dynamics and presented in the local co-ordinate frame of the distal segment. |
| **Joint stiffness** | Ratio between change of joint moments and change in joint angle. |
| **COM power ‡** | (Ground reaction force * COM displacement)/time. |

COM = Centre of mass; ‡ = Manually calculated variable.

### 4.2.6 Statistical Analysis

Minimal sample size for statistical purposes was calculated a priori based on pilot results for vertical whole body stiffness (pre-rehabilitation mean = 0.26, uninjured mean = 0.29, pooled standard deviation = 0.04, power = 0.8, $p<0.05$). The analysis demonstrated that at least 45 subjects would be required per group. Independent t-tests were utilised to compare AGP results to control results pre-rehabilitation and paired t-tests were utilised to compare pre to post changes in the AGP group. No multi comparison adjustment was employed (Perneger, 1998; Hopkins et al., 2009). All results are presented as mean $\pm$ SD. Cohen's effect size was reported as small (0.2 - 0.5), medium (0.5 - 0.8), and large ($> 0.8$) (Cohen, 1988).

## 4.3 Results

### 4.3.1 Subjects

Subject demographics are presented in Table 4.2. Within the AGP group, a primary clinical diagnosis of a pubic aponeurosis injury was made in 46 (71%) cases; hip flexor injury was diagnosed in 14 (22%) cases; adductor injury was diagnosed in 3 (4%) and inguinal injury was diagnosed in 1 (2%) of cases. Patients reported a median time of 36 (IQR 24-73) weeks between onset of symptoms and presentation. Primary sporting participation within both groups was distributed across four

sports with the largest proportion of subjects in both groups playing Gaelic football (Table 4.2).

Table 4.2: Subject demographics and breakdown of primary sporting participation

| Subject demographics | | | | |
|---|---|---|---|---|
| | AGP | | Uninjured controls | |
| Age (yrs.) | 24.6 ± 4.8 (18 – 34.92) | | 23.9 ± 3.4 (20.5 – 30.6) | |
| Height (cm) | 180.5 ± 5.8 (169.0 – 193.5) | | 179.7 ± 9.26 (161.5 – 202.5) | |
| Mass (kg) | 81.5 ± 8.5 (64.3 – 110.1) | | 79.8 ± 13.8 (52.4 – 107.0) | |
| **Sporting participation** | | | | |
| | AGP | | Uninjured controls | |
| | n = | percentage | n = | percentage |
| Gaelic Football | 46 | 70 % | 28 | 56 % |
| Gaelic Hurling | 7 | 11 % | 4 | 8% |
| Soccer | 6 | 10 % | 12 | 23% |
| Rugby | 5 | 9 % | 6 | 13% |

Subject demographics presented as mean ± standard deviation (range).

### 4.3.2 Return to play measures

All AGP subjects completed rehabilitation in a median of 9.14 weeks (IQR 6.6-10.43). The AGP subjects also reported significant improvements in 5 [Pain ($p < 0.01$, $d = 0.83$), Symptoms ($p < 0.01$, $d = 0.75$), Function in daily living ($p < 0.01$, $d = 0.64$), Function in sport and recreation ($p < 0.01$, $d = 1.13$), Quality of life ($p < 0.01$, $d = 1.01$)] out of the 6 subscales following rehabilitation, with only one subsection [Participation in Physical Activities ($p = 0.36$, $d = 0.38$)] not changing significantly.

### 4.3.3 Stiffness Measures

Vertical whole body, ankle plantar flexor, knee extensor and hip abductor stiffness were significantly less in the AGP pre group in comparison to uninjured control group pre-rehabilitation. When the AGP group was compared pre- and post-rehabilitation, hip abductor stiffness increased significantly and ankle internal rotator stiffness decreased significantly. Post-rehabilitation, vertical whole body; ankle plantar flexor and knee extensor stiffness remained significantly less in AGP pre group in comparison to uninjured control group while hip abductor stiffness was no longer significantly different between the two groups (Table 4.3). Hop height was not significantly different between the two groups (AGP pre vs. uninjured control p =0.32, d = 0.15, AGP pre vs. AGP post = 0.43, d = 0.09) nor was hopping width (AGP pre vs. uninjured control: p = 0.13, d = 0.11, AGP pre vs. AGP post p =0.13, d =0.10) and as such were not statistically controlled for in this study.

Table 4.3: Whole body and joint stiffness findings

| Stiffness (*k*) | AGP Pre | AGP Post | Control | AGP Pre vs. Control | | | AGP Pre vs. AGP Post | | | AGP Post vs. Control | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| **Variable** | **Mean ± SD** | **Mean ± SD** | **Mean ± SD** | **Mean Difference** | **D** | **Sig** | **Mean Difference** | **D** | **Sig** | **Mean Difference** | **D** | **Sig** |
| ***Vertical Stiffness*** | 0.24 ± 0.06 | 0.25 ± 0.06 | 0.3 ± 0.07 | 0.06 | 0.79 | 0.00 | 0.01 ± 0.06 | 0.11 | 0.38 | 0.05 | 0.71 | 0.00 |
| **Ankle plantarflexor** | 1.06 ± 0.31 | 1.07 ± 0.32 | 1.25 ± 0.38 | 0.19 | 0.55 | 0.00 | 0.01 ± 0.28 | 0.02 | 0.98 | 0.18 | 0.55 | 0.00 |
| **Hip abductor** | 4.06 ± 9.07 | 7.19 ± 11.29 | 11.45 ± 23.48 | 7.39 | 0.43 | 0.02 | 3.13 ± 12.60 | 0.35 | 0.05 | 4.26 | 0.26 | 0.18 |
| **Knee extensor** | 3.14 ± 1.65 | 2.91 ± 1.45 | 4.05 ± 3.32 | 0.91 | 0.36 | 0.05 | -0.23 ± 1.80 | 0.14 | 0.32 | 1.14 | 0.45 | 0.02 |
| **Knee external rotator** | 0.25 ± 0.27 | 0.32 ± 0.38 | 0.16 ± 0.31 | -0.10 | 0.33 | 0.09 | 0.07 ± 0.30 | 0.26 | 0.06 | -0.17 | 0.50 | 0.01 |
| **Knee abductor** | 17.74 ± 51.19 | 11.15 ± 28.62 | 9.08 ± 20.84 | -8.66 | 0.21 | 0.27 | -6.59 ± 58.92 | 0.13 | 0.37 | -2.06 | 0.13 | 0.49 |
| **Ankle evertor** | 3.19 ± 5.2 | 8.8 ± 30.47 | 2.31 ± 2.69 | -0.88 | 0.20 | 0.29 | 5.60 ± 28.38 | 1.08 | 0.12 | -6.49 | 0.27 | 0.16 |
| **Ankle internal rotator** | 1.99 ± 4.49 | 0.59 ± 1.03 | 1.29 ± 4.25 | -0.70 | 0.16 | 0.41 | -1.40 ± 4.57 | 0.31 | 0.02 | 0.70 | 0.27 | 0.16 |
| **Hip extensor** | 1.46 ± 34.04 | 4.61 ± 22.19 | 11.66 ± 118.52 | 10.20 | 0.13 | 0.51 | 3.15 ± 42.34 | 0.09 | 0.55 | 7.05 | 0.01 | 0.96 |
| **Hip internal rotator** | -2.91 ± 10.54 | -1.29 ± 2.62 | -2.72 ± 15.62 | 0.19 | 0.01 | 0.94 | 1.61 ± 10.83 | 0.15 | 0.24 | -1.42 | 0.08 | 0.72 |

Joint stiffness measured as (Nm.kg-1.degrees-1), vertical stiffness measured as (N.m-1.kg-1), D = Cohen's D effect size, Sig = significance (p). Arranged in order of effect size for AGP pre vs. uninjured.

## 4.4 Discussion

This study investigated if whole body vertical and joint stiffness are affected in subjects with AGP and if so whether return to play following rehabilitation is associated with a change in stiffness in the AGP group. The main finding from this present study was that using a case-control analysis, the AGP group were significantly less stiff in comparison to controls for four of the ten stiffness variables: ankle plantar flexor, knee extensor, hip abductor and whole body vertical stiffness. In contrast, the pre- to post-rehabilitation analysis identified that hip abductor stiffness and ankle internal rotator stiffness changed significantly, while the post-rehabilitation versus uninjured comparison indicated that only hip abductor stiffness was no longer significantly different between the AGP and uninjured group. When examining the biomechanics associated with an injury, previous research has independently reported either case control (injured vs. uninjured) (Delahunt, Monaghan and Caulfield, 2007; Morrissey et al., 2012) or pre- versus post- rehabilitation examinations (Thorp et al., 2010; King et al., 2018). The strength of the current investigation, is that factors identified in the case control analysis that change with rehabilitation (pre- rehabilitation vs. uninjured) and become more similar to the uninjured group (post- rehabilitation vs. uninjured), are more likely to represent true targets for AGP rehabilitation (Marshall and Moran, 2015). The results from this study suggest that hip abduction stiffness may represent a target for rehabilitation. Within this cohort, the AGP group demonstrated a clinical diagnosis across four entities (pubic aponeurosis, hip flexor, adductor and inguinal injury). It could be suggested that these clinical entities would exhibit different movement biomechanics and should be examined individually. However, recent research by our group found no relationship between clinical diagnosis and movement biomechanics (Franklyn-Miller et al., 2017). Further, Franklyn-Miller et al., (2016) contend that AGP is caused by an

overload of the anterior pubic area (pubic symphysis and surrounding tissues), with various structures becoming painful in direct response to this loading or in an attempt to stabilize the region. For this reason we treat AGP as a single entity and have examined the cohort within this current study as a whole group.

### 4.4.1 Pre-rehabilitation differences between the control and AGP groups

Prior to rehabilitation, the AGP group were significantly less stiff in several variables in comparison to controls. This is similar to research by Maquirriain (Maquirriain, 2012) who found that athletes with Achilles tendinopathy presented with reduced whole body vertical stiffness, which the authors' hypothesised was due to increased ankle joint compliance resulting from mechanical and material changes to the Achilles tendon (Arya and Kulig, 2010).

It is plausible that the lower sagittal plane stiffness seen in our study at the ankle, knee and whole body stiffness in the AGP group may represent a compensatory technique. This compensatory technique may reduce loading on the painful pubic symphysis region post injury. The predominantly vertical orientation of both the resultant ground reaction force and stance limb during the hurdle hop action suggests the magnitude of the ground reaction force is associated with whole body and sagittal plane joint stiffness. This resultant ground reaction force passes medially to the hip joint centre during the hurdle hop and will tend to propagate hip adduction. [Whilst the authors acknowledge the projection of ground reaction forces to predict internal joint moments is erroneous (Wells, 1981), we feel this interpretation is useful in understanding the influence of sagittal plane joint stiffness on non-sagittal plane stiffness]. To oppose hip adduction, the ipsilateral hip abductors act eccentrically, producing a concomitant increase in ipsilateral hip joint reaction forces (Neumann, 2010; Wesseling et al., 2015). Any increase in

the force transmitted from the femur to the pelvis may require adjacent muscular, ligamentous, and cartilaginous structures to assist load transfer and may overload the commonly painful pubic symphysis region which is considered the fulcrum around which many forces are exerted at the pelvis (Meyers, Greenleaf and Saad, 2005).

It is also possible that the lower stiffness observed in the AGP group was simply due to reduced neuromuscular capacity, particularly at the ankle and knee. Training volume and intensity is often limited in subjects with AGP, in order to manage the pain (Hölmich and Thorborg, 2014) and neuromuscular detraining can occur in as little as 4 weeks (Neufer et al., 1987). Finally, it is possible that the lower stiffness in the AGP group may reflect a neuromuscular risk factor for AGP with increased joint laxity and strain on localised tissues. Specifically, the lower hip abductor stiffness in the AGP group pre-rehabilitation was associated with an increased range of hip adduction which may result in an increase in hip adductor activity during single leg stance (Prior et al., 2014), and could manifest as increased shear loading at the pubic symphysis (Meyers, Greenleaf and Saad, 2005).

There were six stiffness measures that were not significantly different in the AGP group in comparison to the uninjured controls pre-rehabilitation (knee external rotator, knee abductor, ankle evertor, ankle internal rotator, hip extensor and hip internal rotator) and within the scope of this study, cannot be deemed to be of relevance to AGP. Given the close proximity of the hip joint to the region of pain in AGP patients, it was unexpected that hip extensor stiffness pre-rehabilitation was not lower in the AGP group. A lower hip stiffness would also have reduced the magnitude of resultant ground reaction force, which as indicated above is a potential compensatory mechanism for AGP. While the reason for this finding is unclear, it may be related to the hip acting concentrically to extend during the loading phase of the hurdle hop test (Marshall et al., 2015) and maintain

an upright trunk (Perry, 1992). In order to achieve the task of hopping laterally back over the hurdle (and not hop forward), it would be essential for the athletes in both groups to have an upright trunk at the time take off to avoid forward projection of the body's centre of mass. While some individuals may maintain an upright trunk throughout the hurdle hop since this would be easier to control, others may prioritise absorbing loads at the hip and are comfortable using a large range of motion. These contrasting demands may also explain the large standard deviations in hip extensor stiffness observed in this current study.

### 4.4.2 Changes in stiffness with rehabilitation

Only hip abductor stiffness increased significantly after rehabilitation to become non-significantly different to the uninjured controls. This suggests that in line with previous research (Morrissey et al., 2012), the hip abductors should be targeted in rehabilitation. It is unclear why only one variable increased in stiffness (three further stiffness measures were lower in the AGP group pre-rehabilitation in comparison to the uninjured controls). While this may indicate that only hip abductor stiffness is of relevance to AGP, it could also be argued that the intervention program was ineffective at targeting stiffness qualities (Please see Appendix I for a full description of the exercise intervention.) . Indeed, while plyometric exercises are an effective means of enhancing whole body and sagittal plane stiffness (Brazier, Bishop and Simons, 2014), the volume and intensity of plyometric exercises included within this study was less than previously reported (Toumi et al., 2004; Kubo et al., 2007; Cormie, McGuigan and Newton, 2010). Furthermore, much of the rehabilitation is focused on hip extensor and lateral hip strength, aimed to enhance hip abductor function, and may have done so preferentially. These findings may indicate that an intervention with a greater emphasis on increasing joint stiffness may be warranted. However, care should

be taken, as if stiffness represents a compensatory mechanism or neuromuscular detraining in AGP, then increasing stiffness in AGP subjects to normative ranges may increase loading on the painful pubic symphysis region. It is the opinion of the authors however, that a lack of training stimulus may not be the sole explanation for the lack of change in stiffness. Firstly, the training intervention included multiple training modalities (e.g. strength, power, plyometrics) that have been shown to enhance stiffness, and secondly all AGP patients in this study returned to pain-free participation in sport, suggesting that whole body, knee and ankle stiffness are not of relevance to AGP rehabilitation. Future research should prospectively track AGP patients following return to play to determine if stiffness is related to reoccurrence of this condition and also determine if an intervention with a greater focus on increasing whole body, knee and ankle stiffness improves the efficacy of AGP rehabilitation.

### 4.4.3 The challenges of measuring stiffness

Generally when examining joint stiffness, the loading phase of a moment-angle graph is assumed to be a clear biphasic pattern (Farley and Morgenroth, 1999; Kuitunen, Komi and Kyrölinen, 2002; Hamill, Gruber and Derrick, 2014). However, biphasic patterns are not always present, with fluctuating moments [between agonist and antagonist muscle dominance (Figure 4.2)] occurring even within the sagittal plane at the knee (Bezodis, Kerwin and Salo, 2008; Marshall et al., 2015) and hip (Bezodis, Kerwin and Salo, 2008; Muraki et al., 2008; Schache et al., 2011; Bencke et al., 2013; Marshall et al., 2015) during sprinting, jumping, hopping and cutting. This results in polyphasic patterns, where we believe it is inappropriate to examine a single stiffness measure across the entire waveform (see methods). Within this study, many trials produced no eccentric stiffness (e.g. 26% of the hip extensor trials), which traditionally, if included in stiffness calculations, would

lead to erroneous findings. Additionally, even when examining stiffness of a biphasic pattern, net joint moments are often included that are dominated by agonist and antagonist muscle groups (e.g. knee extensor and flexors) in a single stiffness measure (Farley and Morgenroth, 1999; Hamill, Moses and Seay, 2009; Ford, Myer and Hewett, 2010; Sinclair et al., 2015). This is problematic when the moment-angle gradient is non-linear and suggests the need to screen waveforms and where required utilise alternative stiffness measures. Within this study, a novel stiffness calculation was presented which accounts for varying moment dominance and discontinuous regions of interest.

### 4.4.4 Limitations

This study examined a lateral hopping action. This provided a more ecologically valid examination of stiffness in field sport athletes in comparison to vertical hopping. However, it is not known if equivocal results would be obtained if stiffness were examined during a predominantly sagittal motion, as is the norm within the literature (Farley et al., 1998; Chelly and Denis, 2001; Brauner et al., 2014). To date, within the biomechanics literature, stiffness has been examined as a discrete value (e.g. instantaneous stiffness at a specific time point, or the average stiffness between two time points). Using a discrete value for stiffness however, may misrepresent localised periods of high or low stiffness at various time points during the landing phase. To address this limitation, future research should explore continuous signal joint stiffness measures.

Progression through the intervention in this present study was dependent on achieving set criteria. Whilst this is a more ecologically valid approach to rehabilitation than time-based progression, it is important to note that the variation in reassessment time post-rehabilitation may be a confounding factor when examining how change in stiffness is associated with return to play in this cohort

cohort as it is possible that the change in stiffness is related to the presence of pain and not to potential propagative movements.

Given the focus of this paper on identifying if stiffness is associated with athletic groin pain, only patients who completed the exercise intervention and returned for post intervention testing were included in this study. As such, potentially important information regarding the mechanics of those who do not successfully rehabilitate is not gained from this study. Finally, given the retrospective nature of this study, it is unclear if the pre-rehabilitation mechanics exhibited by the AGP group are a cause or result of the injury itself. This is particularly pertinent with respect to stiffness given the changes that pain can elicit at a central nervous system level (Kotler et al., 1998; Hodgesv and Tucker, 2011) and the dominant role the central nervous system plays with respect to modulating stiffness. Prospective research is therefore required to clearly ascertain if stiffness is a risk factor for AGP.

## 4.5 Conclusion

This was the first study to investigate stiffness in AGP patients. While a causal relationship cannot be investigated within the current study design, it would appear that ankle plantar flexor, knee extensor, hip abductor and whole body vertical stiffness is affected by AGP and with the exception of hip abductor stiffness, do not improve following clearance to return to play. These findings suggest that hip abductor stiffness may represent a target for rehabilitation in AGP patients. Conversely, it is likely that the lower sagittal plane and whole body vertical stiffness in the AGP group in comparison to the uninjured controls represents either a compensatory mechanism to reduce the peak magnitude of the hip joint reaction force or is a reflection of detraining. As a result of this study the authors now believe that hip mechanics in the frontal plane are of particular

relevance to AGP and hip abductor stiffness has been further emphasised within the AGP rehabilitation programme at the Sports Surgery Clinic. The authors are currently in the process of investigating if rehabilitation with greater focus on increasing whole body, knee and ankle stiffness improves the efficacy of AGP rehabilitation. Future research is also warranted to prospectively track AGP patients following return to play to conclusively determine if stiffness is related to reoccurrence of this condition. To avoid potentially erroneous findings, future researchers examining stiffness should screen moment-angle waveforms and where required utilise alternative stiffness measures, such as presented within this study.

### 4.5.1 Link between Chapter 4 and Chapter 5

Chapter 4 sought to investigate if stiffness is affected by AGP. The results from this study suggest that hip abductor stiffness may represent a target for rehabilitation in AGP patients.

While Chapter 3 and Chapter 4 utilised different measures to explore AGP, both measures were in the magnitude time domain. Another potentially important source of information is the structure of a signal, for example; its complexity which is defined as the deterministic structural richness contained within a signal (Grassberger, 1991). It has been suggested that complexity is reduced with pathology (Harbourne and Stergiou, 2009), yet to date, no research has investigated complexity with respect to AGP. The purpose of Chapter 5 was to investigate the utility of complexity to delineate between those with and without AGP.

# Chapter 5

# Biomechnical Complexity: A useful measure to delineate between those with and without athletic groin pain?

## 5.1 Introduction

Biomechanical analysis of injuries (including AGP) has traditionally focused on magnitude-based representation (e.g. maxima and minima values, standard deviation, time to event etc.) of biomechanical signals (e.g. joint moment and angles) using discrete (Marshall and Moran, 2015) or continuous signal analysis (Pataky, 2010; Richter, O'Connor, et al., 2014b). However, the structure of a signal, for example its complexity, is also a rich source of additional information (Pincus and Huang, 1991; Richman and Moorman, 2000). Complexity refers to the deterministic structural richness contained within a signal (Grassberger, 1991) that emerges from the dynamic interaction of multiple components, organized around and summating to an outcome goal (De Rosnay 1975 cited Komar et al. 2015) (e.g. lateral hurdle hops for speed). Unlike inter-trial variability, which simply quantifies the magnitude of deviation from a measure of central tendency,

complexity investigates the order and regularity within a signal. Biological systems are complex in nature, and it is this complexity that is thought to reflect the system's capacity to respond to a constantly changing environment (Goldberger et al., 2002; Lipsitz, 2002). Pathology or injury is theorized to reduce the degrees of freedom available to the system to achieve a movement task and hence there may be a reduction in signal complexity (Harbourne and Stergiou, 2009).

While there are a number of measures of complexity, including 'detrended fluctuation analysis' and 'lyapunov exponent' one of the best-known methods of quantifying signal complexity is entropy, and as such has been utilised in a variety of disciplines including cardiology, neurology, motor control and biomechanics (Cavanaugh et al., 2006; Di, Industriale and Informazione, 2014; Terada et al., 2015). Derived from information theory, the underlying principle of entropy is that simple sequences can be described concisely whilst complex sequences cannot. Greater entropy is therefore associated with more complexity. A number of entropy methods have been proposed to measure complexity of a signal, including approximate entropy (Pincus and Huang, 1991) and sample entropy (Richman and Moorman, 2000). However, both of these methods are sensitive to data length, in that very short data signals will reduce the discriminative power of both these methods (Yentes, Hunt and Schmid, 2013).This requirement for a long data signal is common to large number of complexity measures not just entropy (Stergiou, 2016) which makes them unsuitable for exploring discrete, non-cyclic movement tasks such as the hurdle hop in this study.

To overcome this requirement for a long data signal, a recently devised metric termed quadratic sample entropy (QSE) was proposed by Lake and Moorman (2011) which is robust to signal length. QSE is an evolution of sample entropy and has been utilised in a number of domains such as the detection of atrial fibrillation (Lake and Moorman, 2011), the classification of electroencephalograms to detect

Alzheimer's disease (Simons, Abasolo and Escudero, 2015) and the analysis of heart rate variability (Liu et al., 2013).

Although not commonly applied in human motion or musculoskeletal injury research, emerging literature suggests that measures of complexity may be more sensitive in detecting differences between movement signals compared to traditional methods, as traditional methods do not account for the non-linear dynamics of a signal (Stergiou and Decker, 2011). Athletic groin pain (AGP) is a prevalent musculoskeletal injury in sports that involve rapid changes in direction whilst running at high speeds (Murphy et al., 2012; Thorborg, Rathleff and Petersen, 2015). Whilst recent research has begun to examine the biomechanics of AGP using traditional metrics (Franklyn-Miller et al., 2017; Janse van Rensburg et al., 2017; Severin et al., 2017; King et al., 2018), examinations of stiffness (Gore et al. 2018, [Chapter 4]) and measures of the amount of linear variability (Edwards, Brooke and Cook, 2017), to date no research has examined complexity in AGP patients and its ability to delineate between injured and uninjured groups. It has been suggested that inappropriate loading during multi-planar exercise is a source of AGP (Franklyn-Miller et al., 2017). As such it may be most appropriate to examine complexity in AGP patients across multiple planes during a dynamic multi-planar movement such as the lateral hurdle hop (Gore et al. 2018, [Chapter 4]).

Based on the emerging method of QSE, the purpose of this study is to examine the complexity of moment, angle and angular velocity waveforms in both AGP and uninjured populations during a lateral hurdle hop task. It was hypothesised that the AGP group would have significantly lower complexity when compared to uninjured group.

## 5.2 Methods

Ninety six male subjects (mean ± SD: age 24.9 ± 5.9 yrs., height 180.2 ± 6.0 cm, mass 81.0 ± 9.5 kg) attending the Sports Surgery Clinic, Dublin, along with 50 male controls (mean ± SD: age 23.9 ± 3.4 yrs., height 179.7 ± 9.26 cm, mass 79.8 ± 13.8 kg) were recruited. Inclusion criteria for this study included being involved in field-based sports and between the ages of 18 - 35. AGP patients were required to have been clinically diagnosed with AGP as outlined in a previous study (Falvey, King and Kinsella, 2015), and the Controls were required to be injury free for at least two weeks prior to testing with no history of AGP. The uninjured group were matched to the AGP group based on sport, participation level, age and leg dominance. Ethical approval for this study was granted by the Sports Surgery Clinic's ethics committee and all of the participants signed informed consent. The study was registered at Clinicaltrials.gov (NCT02437942).

### 5.2.1 Measurements and Protocols

Each participant attended the biomechanics laboratory on one occasion. Prior to the experimental testing, subject mass and height was recorded using an electronic scale (Seca 876) and stadiometer (Seca 213) and each subject completed a standardised dynamic warm-up involving a three-minute treadmill jog at 8 km/h, five body weight squats and five practice trials of the hurdle hop test. The subjects had no previous experience of the hurdle hop test. The hurdle hop involved the participants hopping laterally over a 15 cm hurdle followed by an immediate hop back to the initial starting position. Participants were instructed to undertake the hop as explosively as possible and it was the first initial landing phase that was analysed. The distance between foot contacts was 40 cm (the distance between force plate centres). Three trials were completed to generate a mean complexity score. The AGP participants began on their symptomatic leg, contralateral non-

weight bearing foot behind with the knee flexed to approximately 90 degrees, and hands unrestricted for balance (Figure 5.1). Similar to Franklyn Miller et al., (2016), when the AGP patient had bilateral symptoms, the leg examined was chosen at random. The uninjured group was proportionally matched accordingly based on leg dominance.

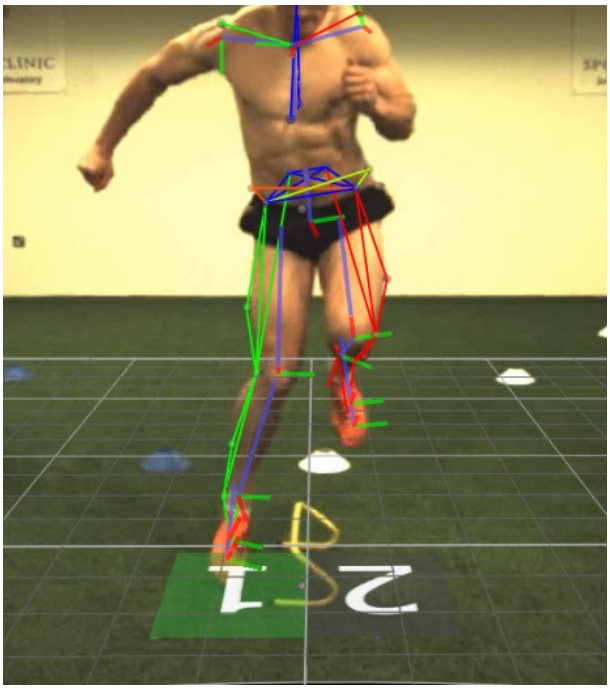

Figure 5.1: Hurdle Hop exercise depicting the initial landing of interest

### 5.2.2 Data Capture

Twenty-four Reflective markers (14 mm diameter) were placed at bony landmarks on the lower limbs, pelvis and trunk as per the Vicon Plug in Gait model (Vicon Motion Systems, Oxford, UK) (Marshall et al., 2014). Three-dimensional marker motion was tracked using 8 infrared cameras (Vicon - Bonita B10, UK) sampling at 200 Hz. This was synchronized with two 40x60 cm force platforms (AMTI - BP400600, USA) collecting ground reaction force data at 1000 Hz.

### 5.2.3 Data Analysis

Both marker and force data were filtered using a fourth order Butterworth filter with a cut-off frequency of 15 Hz (Kristianslund et al. 2013). Moment and angle calculations were performed in Nexus software (Vicon Motion Systems, Oxford, UK). The data were subsequently exported to Matlab 2013b (Mathworks, USA) where angular velocity and signal entropy were calculated and the statistical analyses conducted.

To calculate signal entropy, the period of initial contact to toe off was examined. Initial contact and toe off were defined respectively as the instances at which the vertical ground reaction force exceeded and fell below a 5N threshold. No time normalisation procedures were performed to avoid any kind of alteration to the dynamics of time series. Entropy was calculated using quadratic sample entropy as outlined in Lake and Moorman (2011). The underlying concept of entropy is that in simple waveforms, sequences or subsections are repeated regularly, while this is not the case in a complex waveform. In this respect, the first step of calculating quadratic sample entropy is to calculate the conditional probability that two short vectors or 'subsections' of a waveform that match within a tolerance of acceptability will continue to match at the next point. This calculation is termed sample entropy and is calculated as follows: given a waveform containing $t$ consecutive data points ($x_1, x_2, x_3 ... x_t$), a vector or 'subsection' of this waveform of length $m < t$ and starting at point $i(x_i, x_{i+1} ... x_{i+m-1})$ is termed a template [$Temp_m(i)$]. This template is compared to subsequent vectors or 'subsections' of length m within the total waveform. Every time a match between the template and subsequent subsections of the waveform is observed within an acceptable tolerance of dissimilarity (r >0) (The template vector is compared to subsequent vectors in the same waveform. If the template vector is within a pre-specified distance (r) from subsequent vectors in that waveform it is termed a match.), the

match is counted and their conditional probabilities are summed and divided by $t - m$. This creates the variable A. This process is then repeated for [$Temp_{m+1}(i)$] creating the variable B. The sample entropy (Equation 5.1) is then calculated as the negative natural logarithm of the conditional probability of a match of length $m + 1$ given a match of length $m$:

$$Sample\ Entropy = -ln(\frac{A}{B}) \tag{5.1}$$

Quadratic sample entropy can then be calculated by adding the natural logarithm of $2r$, thereby removing the influence of the size of $t$ through normalisation (Lake and Moorman, 2011) (Equation 5.2):

$$Quadratic\ Sample\ Entropy = SampleEntropy + ln(2r) \tag{5.2}$$

For this study QSE was calculated with $m = 2$ and $r = 0.2*$ standard deviation of the signal being examined (Pincus and Huang, 1992; Yentes, Hunt and Schmid, 2013). A series of t-tests were utilised to compare AGP results to uninjured results however no adjustment was deemed necessary (Perneger, 1998; Hopkins et al., 2009). All results are presented as mean $\pm$ SD. Cohen's effect size was reported as small (0.2 - 0.5), medium (0.5 - 0.8), and large ($>$ 0.8) (Cohen, 1988). A summary of the biomechanical variables examined using QSE in this chapter is presented in Table 5.1.

Table 5.1: Summary of Biomechanical variables examined using quadratic sample entropy

| Variable | Description of Calculation |
|---|---|
| **Ankle, Knee and Hip joint angle** | Relevant angle between two adjacent segments described with respect to the proximal coordinate system. |
| **Thorax and Pelvis angle** | Absolute angle relative to global coordinate system. |
| **Joint moments** | Calculated using inverse dynamics and presented in the local co-ordinate frame of the distal segment. |
| **Joint/ Segment velocity ǂ** | Derivative of joint/ segment angle |

COM = Centre of mass; ‡ = Manually calculated variable.

## 5.3 Results

Of the 15 complexity results for the angle waveforms (Table 5.2) only hip ab/adduction was significantly greater in the AGP group in comparison the uninjured control group (effect size: 0.44). For the complexity of the angular velocity findings (Table 5.3), 5 of the 15 variables were significantly less in AGP group (effect size: 0.36-0.49). For the moment waveforms (Table 5.4), 7 of the 9 complexity results were significantly less in AGP group with effect sizes ranging from (0.31-0.96).

Table 5.2: Mean quadratic sample entropy for angle waveforms (ordered in terms of effect size)

| Variable | AGP | Control | AGP Pre vs. Control | |
|---|---|---|---|---|
| | Mean ± SD | Mean ± SD | D | Sig |
| Hip Ab/Adduction | 0.58±0.38 | 0.42±0.33 | 0.44 | 0.01 |
| Plevis Int/Ext Rotation | 0.17±0.34 | 0.07±0.36 | 0.30 | 0.09 |
| Knee Ab/Adduction | 0.39±0.36 | 0.5±0.38 | 0.29 | 0.11 |
| Pelvis Flex/Ext | -0.04±0.32 | 0.04±0.25 | 0.27 | 0.13 |
| Pelvis Ab/Adduction | 0.34±0.37 | 0.27±0.31 | 0.22 | 0.22 |
| Knee Flex/Extension | 1.66±0.18 | 1.63±0.2 | 0.18 | 0.31 |
| Thorax Int/Ext Rotation | 0.2±0.36 | 0.13±0.4 | 0.18 | 0.31 |
| Ankle Ab/Adduction | -0.76±0.54 | -0.84±0.5 | 0.15 | 0.39 |
| Ankle Dor/Plan flexion | 1.84±0.16 | 1.86±0.13 | 0.14 | 0.42 |
| Hip Int/Ext Rotation | 0.82±0.35 | 0.87±0.37 | 0.14 | 0.43 |
| Thorax Flex/Ext | -0.04±0.38 | 0.00±0.36 | 0.12 | 0.52 |
| Thorax Ab/Adduction | -0.42±0.35 | -0.44±0.31 | 0.06 | 0.74 |
| Ankle Int/Ext Rotation | 1.18±0.31 | 1.19±0.31 | 0.04 | 0.84 |
| Hip Flex/Extension | 1.04±0.2 | 1.03±0.26 | 0.02 | 0.90 |
| Knee Int/Ext Rotation | 1.26±0.29 | 1.26±0.26 | 0.00 | 1.00 |

Dor-Plan = Dorsi-plantar, Int/Ext = Internal-external, D = Cohen's D effect size, Sig = significance (p).

Table 5.3: Mean quadratic sample entropy for angular velocity waveforms (ordered in terms of effect size)

| Variable | AGP Pre | Control | AGP Pre vs. Control | |
|---|---|---|---|---|
| | Mean ± SD | Mean ± SD | D | Sig |
| Ankle Dor/Plan flexion | 2.9±0.25 | 3.01±0.18 | 0.49 | 0.01 |
| Hip Flex/Extension | 2.22±0.24 | 2.32±0.25 | 0.41 | 0.02 |
| Hip Int/Ext Rotation | 2.41±0.56 | 2.62±0.49 | 0.39 | 0.03 |
| Thorax Flex/Ext | 0.87±0.34 | 1.01±0.35 | 0.38 | 0.03 |
| Knee Flex/Extension | 2.89±0.22 | 2.96±0.19 | 0.36 | 0.04 |
| Knee Ab/Adduction | 2.12±0.51 | 2.28±0.52 | 0.32 | 0.07 |
| Thorax Ab/Adduction | 0.54±0.37 | 0.64±0.27 | 0.30 | 0.09 |
| Thorax Int/Ext Rotation | 0.72±0.28 | 0.79±0.27 | 0.24 | 0.17 |
| Knee Int/Ext Rotation | 2.83±0.53 | 2.95±0.44 | 0.23 | 0.20 |
| Ankle Int/Ext Rotation | 2.71±0.48 | 2.82±0.46 | 0.23 | 0.20 |
| Pelvis Flex/Ext | 1.09±0.41 | 1.15±0.35 | 0.17 | 0.35 |
| Pelvis Ab/Adduction | 1.3±0.37 | 1.34±0.28 | 0.11 | 0.53 |
| Hip Ab/Adduction | 1.59±0.35 | 1.55±0.34 | 0.11 | 0.55 |
| Plevis Int/Ext Rotation | 1.17±0.34 | 1.2±0.34 | 0.10 | 0.58 |
| Ankle Ab/Adduction | 0.77±0.68 | 0.8±0.66 | 0.04 | 0.82 |

Dor-Plan = Dorsi-plantar, Int/Ext = Internal-external, D = Cohen's D effect size, Sig = significance (p).

Table 5.4: Mean quadratic sample entropy for moment waveforms (ordered in terms of effect size)

| Variable | AGP Pre | Control | AGP Pre vs. Control | |
|---|---|---|---|---|
| | Mean ± SD | Mean ± SD | D | Sig |
| Hip Flex/Extensor | -0.50±0.45 | -0.05±0.37 | 0.96 | >0.01 |
| Ankle Dor/Plan flexor | -0.68±0.28 | -0.47±0.27 | 0.74 | >0.01 |
| Hip Ab/Adductor | -0.97±0.30 | -0.75±0.34 | 0.67 | >0.01 |
| Knee Flex/Extensor | -0.33±0.26 | -0.18±0.23 | 0.62 | >0.01 |
| Hip Int/Ext Rotator | -2.20±0.32 | -2.03±0.32 | 0.53 | 0.01 |
| Knee Int/Ext Rotator | -2.62±0.46 | -2.39±0.42 | 0.49 | 0.01 |
| Ankle Int/Ext Rotator | -2.12±0.39 | -1.94±0.40 | 0.43 | 0.02 |
| Ankle Ab/Adductor | -2.89±0.52 | -2.72±0.53 | 0.31 | 0.10 |
| Knee Ab/Adductor | -0.84±0.35 | -0.73±0.40 | 0.31 | 0.11 |

Dor-Plan = Dorsi-plantar, Int/Ext = Internal-external, D = Cohen's D effect size, Sig = significance (p).

## 5.4 Discussion

This exploratory study sought to investigate if biomechanical complexity was a useful measure to delineate between those with and without AGP during a lateral hurdle hop task. The overall findings of this investigation identified that the AGP group had significantly lower complexity than the uninjured group for 78% of the moment variables and 33% of the angular velocity variables examined.

In contrast, only 7% of the variables were significantly different for joint angles, and were opposite to the moment and angular velocity results in that the AGP group demonstrated greater signal complexity. It is unclear why the findings of complexity differed for joint angles, but it possibly reflects: (i) joint moment and angular velocity signals being more complex per se, and (ii) the higher degree of freedom in measures of movement control (e.g. joint moments) than movement outcome (e.g. joint angles) (Bernstein, 1967; Winter et al., 1990). Given that the largest effects sizes were evident for the moment findings (d=0.31-0.96), the following discussion will focus on moment complexity.

The lower joint moment complexity observed in the AGP group is supported by previous research indicating that injured individuals with chronic ankle instability (Terada et al. 2015) and knee osteoarthritis (Tochigi et al. 2012) have lower complexity in measures of ankle frontal plane kinematics and tri-axial leg accelerations, respectively, during walking. In other domains, lower signal complexity has also been identified in centre of pressure data in concussed athletes (Cavanaugh et al., 2006) and in various pathological biological signals including: heart rate atrial fibrillation (Lake and Moorman 2011), electrocardiographic measures of atrial fibrillation (Alcaraz and Rieta, 2010) and electroencephalograms in Alzheimer's disease (Simons et al. 2015).

While the present study did not examine the underlying pathophysiology associated with this loss of complexity in AGP patients, it has been suggested within the general biological literature that reduced complexity is associated with a reduction in the number of, or the coupling and co-ordination between, sensory inputs (Pincus, 1994; Newell, 1998; Cavanaugh, Guskiewicz and Stergiou, 2005). For example research examining the effects of eyes open vs. eyes closed conditions during quiet standing has demonstrated that the eyes closed condition reduces the complexity of postural sway (Ramdani et al., 2009). In accordance with this concept, it is possible that reduced complexity in AGP patients is due

to interference within and/or between systems of the body. This manifests in a reduction in the degrees of freedom available to the AGP group during the execution of the hurdle hop action and possibly a reduced capacity to rapidly respond to perturbations that arise during the landing phase of this task (Cavanaugh, Guskiewicz and Stergiou, 2005). There are a number of hypotheses as to why the number of, or the coupling between, sensory inputs is reduced within the AGP group in this present study. Firstly, the lower complexity in the AGP group may reflect a risk factor for the development of this condition. It is possible that the reduced complexity in the AGP group during the execution of dynamic loading tasks results in an inability to respond to perturbations that commonly occur during field-based sports. It is unclear however whether reduced complexity precedes AGP or is a result of this injury, and further research is warranted.

Secondly, it is possible that the AGP group are utilising a more regular, rigid, motor behaviour in an attempt to avoid pain (or perceived threat of pain) associated with this condition. Indeed while it is well accepted that pain is a potent stimulus to alter technique, research has also demonstrated that compensatory movement can be retained when pain is no longer present (Moseley and Hodges, 2006; Tucker et al., 2012). It has been suggested that these compensatory adaptions to pain involve changes at multiple levels of the motor system and leads to the redistribution of activity within and between muscles (Hodgesv and Tucker, 2011). The association between pain and complexity is illustrated by recent investigations into seated postural discomfort and complexity (Sondergaard et al., 2010) indicating that discomfort [an early perception of pain (Madeleine et al., 1998)] is associated with reduced complexity of postural control. Compensatory techniques are common in various injuries including AGP (Gore et al. 2018, [Chapter 4]) and anterior cruciate ligament injury (Paterno et al., 2010). However, while constrained (lower in complexity) movement patterns may serve to reduce the

risk of injury or pain in the short term, they can often predispose to other injuries (Davis and Seol 2005) and as such are not advised as long-term solutions.

A final explanation of our findings is that the lower complexity may reflect neuromuscular detraining. Indeed previous research has demonstrated that heart rate complexity is reduced after just four weeks of detraining (Heffernan et al., 2007) and a reduction of training load is common in AGP in order to manage the pain associated with this condition (Hölmich and Thorborg, 2014).

Interestingly, when exploring the moment results of the present study (Table 5.4), there appears to be two patterns of findings in relation to the dominance of effect sizes. Firstly, of the three joints examined, the hip joint in general demonstrated the largest effect sizes when comparing the AGP and the uninjured group. This is perhaps not surprising given previous research has postulated that alterations in hip mechanics may be involved in the pathomechanics of groin pain (Franklyn-Miller et al. 2017, Gore et al. 2018, King et al. 2018, Janse van Rensburg et al. 2017, Severin et al. 2017). During dynamic weight bearing tasks, the hip joint works to transfer load from the lower limb to pelvis. Any alteration in hip mechanics will therefore not only alter the resultant hip joint force (Neumann, 2010; Wesseling et al., 2015), but also overload the commonly painful pubic symphysis region and the adjacent muscular, ligamentous, and cartilaginous structures, which act to stabilise it (Meyers, Greenleaf and Saad, 2005). Indeed hip pathologies are a frequently concurrent presentation with AGP (Hiti et al., 2011). Secondly, of the three planes examined, the sagittal plane moments were best able to discern between the AGP group and the uninjured group as determined by effect sizes. This finding is supported by our previous research on another AGP cohort using traditional linear measures (Gore et al. 2018, [Chapter 4]), which suggests that increased sagittal plane loading during single leg tasks could increase the magnitude of resultant ground reaction force passing medial to the

hip joint resulting in greater hip abductor moments and a concomitant increase in hip joint reaction force (Neumann, 1989).

Rehabilitation is a central component of injury management. In light of the findings within this present study, future research should explore the need to target rehabilitation techniques that normalise movement complexity in AGP patients to uninjured magnitudes. While traditional methods of rehabilitation (e.g. resistance and balance training) appear to improve (appropriately increase) complexity of various physiological systems (Heffernan et al., 2007; Wayne and Kaptchuk, 2008; Millar and Levy, 2013), it is possible that interventions aiming to optimise the restoration of biological complexity should follow a dynamic systems theory approach (Newell 1986). Such an approach would prioritise variable practice, target multiple systems and facilitate the emergence of functional movement solutions through the interaction of patient, task and environment constraints (Harbourne and Stergiou, 2009; Lee et al., 2014). Rehabilitation may also benefit from emphasising the use of a large number of degrees of freedom to allow the re-instatement of complexity in movement patterns. Future research is warranted to determine if reduced complexity can be used to screen for predisposition to injury and to examine the effects of different rehabilitations strategies on joint moment complexity.

Finally, given that complexity potentially represents the underlying pathophysiology that occurs with injury (Harbourne and Stergiou, 2009), future research should also determine if complexity could be used as a more effective outcome measure following rehabilitation than currently utilised methods. This is particularly pertinent in AGP given the relatively high rate of re-injury (Hagglund, 2006).

### 5.4.1 Limitations

While there are a number of techniques that can be utilised to examine complexity, within this study only entropy was utilised. Specifically, QSE was examined given its suitability to short data signals as examined in this study (Lake and Moorman, 2011). It is acknowledged however, that reliance on any single test may give a misleading representation of physiological complexity (Goldberger 2002). Future research should therefore explore additional suitable measures of complexity to examine in AGP.

## 5.5 Conclusion

The results from this exploratory study suggest that biomechanical complexity is a useful measure to delineate between those with and without athletic groin pain. Within this study, AGP patients were characterised by lower complexity in measures of joint moments and joint angular velocities in comparison to an uninjured group. Based on effect size, moments were the most sensitive to discerning between the AGP and controls. Within the moment results, the hip joint could distinguish the two groups better than the ankle and knee joint, whilst of the three planes examined the sagittal plane was most sensitive to discerning between the AGP and controls. These findings suggest that biomechanical complexity can distinguish between those with and without an injury. Furthermore, complexity may be useful as a rehabilitation outcome measure, but this requires investigation. While the underlying pathophysiology of reduced complexity in AGP is unclear, our findings may suggest that rehabilitation might be guided by a dynamic systems theory approach, but this also requires specific investigation.

### 5.5.1 Link between Chapter 5 and Chapter 6

Chapter 5 sought to compare the complexity of joint angle, angular velocity and moment waveforms between AGP patients and uninjured controls. The results from this study suggest that biomechanical complexity is a useful measure to delineate between those with and without AGP. In fact, for the comparison of those with and without AGP, Chapter 5 generally demonstrated greater effect sizes in comparison to Chapter 3 and 4 which also examined the hurdle hop task.

For the three chapters investigating the hurdle hop task (Chapter 3 , 4 and 5), a mean of three trials was calculated as a representative movement pattern. While taking the mean of three trials is typical within the biomechanical literature (Mullineaux, Bartlett and Bennett, 2001), the scientific justification for doing so is lacking. The aim of Chapter 6 was to experimentally identify the number of trials required for a representative mean of joint angle and moment values during a lateral hurdle hop test.

# Chapter 6

# The number of trials required to obtain a representative movement pattern during a hurdle hop exercise

This study has previously been published in full:

Gore, S.J., Marshall, B.M., Franklyn-Miller, A.D., Falvey, E.C. and Moran, K.A., 2016. The Number of Trials Required to Obtain a Representative Movement Pattern During a Hurdle Hop Exercise. Journal of applied biomechanics, 32(3), pp.295-300

It is presented here in this chapter with only minor changes to confirm to the style and formatting of this thesis.

## 6.1 Introduction

The lateral single leg hurdle hop is a novel biomechanical screening exercise proposed as part of a testing battery for athletic groin pain patients (Marshall et al., 2015, 2016). Both single leg consecutive hopping and non-consecutive hops have been used as measures of stiffness (Brauner et al., 2014) and function in rehabilitation (Myer et al., 2006). Unlike stationary or forward hopping however, the lateral hurdle hop may be a more effective means of stressing frontal plane control patterns typical of field based sports (Newman, Tarpenning and Marino, 2004) and its non-consecutive nature may also have greater ecological validity to sporting actions in comparison to consecutive hopping (Maloney, Fletcher and Richards, 2015). While previous research (Myer et al., 2005; Ageberg et al., 2008) has reported the results of a single hopping trial, the natural variability associated with human movement (Bartlett, Wheat and Robins, 2007) would suggest that a single trial may not be representative, leading to erroneous or at least incomplete findings. Indeed within any single measurement, the observed result ($X_o$) is a summation of the true result ($X_t$) and an inconstant component ($i$) that represents the natural human variability (DeVellis, 2006).

$$X_o = X_t + i \tag{6.1}$$

As a practical solution to this, biomechanical research has traditionally utilized the mean of a small number of trials, typically three (Mullineaux, Bartlett and Bennett, 2001), but also ranging from one to six trials (Rudolph, Axe and Snyder-Mackler, 2000; Webster, Gonzalez-Adrio and Feller, 2004; Myer et al., 2005; Augustsson et al., 2006; Orishimo et al., 2010). Clearly however, the scientific justification of these numbers is lacking. Classical test theory would suggest that the mean of a subject's observed results across multiple trials will asymptote to the true result as the trial number reaches infinity (Beckstead, 2013)

$$\bar{X}_o = X_t \tag{6.2}$$

While increasing the number of trials will improve statistical power and reliability (Mullineaux, Bartlett and Bennett, 2001) there is a trade-off between the number of trials required and the practicality of data acquisition. For this reason, researchers have determined statistically how many trials are required to adequately represent a participant's performance (Bates et al., 1983; Rodano and Squadrone, 2002; James et al., 2007; Racic, Pavic and Brownjohn, 2009).

Within the literature, there are two main methods of determining how many trials are required. Utilizing a intra class correlation coefficient ($ICC_{3,1}$) approach, the number of trials required is determined when the $ICC_{3,1}$ reaches its peak value ($ICC_p$) (James et al., 2007). By contrast sequential analysis methods utilize a cumulative mean and a predefined bandwidth of precision where a representative mean is achieved once the cumulative mean falls within the bandwidth and remains there for all subsequent trials (Hamill and McNiven, 1990).

The aforementioned methods are not without their limitations. The $ICC_p$ method is subject to falsely defining performance stability at two trials should these trials by chance be highly correlated. Indeed, in a study of jumping using this method, the authors (Racic, Pavic and Brownjohn, 2009) observed that many variables achieved stability after only two consecutive trials. The sequential analysis method is limited by its use of an arbitrarily defined bandwidth of precision [typically 25% (James et al., 2007; Racic, Pavic and Brownjohn, 2009) or 30% (Rodano and Squadrone, 2002) of the within subject standard deviation].

A potential solution to these limitations is the use of a novel sequential analysis approach coupled with a bandwidth defined by a modified version of the standard error of measurement (SEM) (see methods). The theoretical grounding of this method is that on an individual level the SEM has been utilized as a measure

of change (Hopkins, 2000; Witmer, Davis and Moir, 2010; Healy and Harrison, 2014) whereby values outside 1.5 times the SEM would indicate that a true change had occurred. In this respect when the cumulative mean stays within the SEM bandwidth it can be said to not represent a true change, that is, it is representative.

The aim of this study is to identify and compare the number of trials required to achieve a representative mean during the lateral hurdle hop exercise using the $ICC_p$ method and the sequential analysis method at three bandwidths: 25% ($SD_{25}$) and 30% ($SD_{30}$) of the within subject standard deviation, and a bandwidth defined by a modified version of the SEM ($SEM_{ind}$).

## 6.2 Methods

### 6.2.1 Participants

15 recreationally active male field sport athletes (mean $\pm$ SD: age 25.85 $\pm$ 3.3 yrs., height 1.77 $\pm$ 0.79 m, weight 77.37 $\pm$ 10.7 kg) were recruited for this study. Ethical approval was granted by the Sports Surgery Clinic ethics committee and all participants signed informed consent.

### 6.2.2 Experimental Protocol

The participants completed 15 trials for a single leg hurdle hop. Prior to the experimental testing, the subjects completed a standardized warm-up similar to (Marshall et al., 2015). This involved a three-minute treadmill jog at 8 km/h, five body weight squats and five practice trials of the hurdle hop test. The subjects had no other experience of the hurdle hop test. The test involved a lateral hop over a 15cm hurdle followed by an immediate hop back to the initial starting position. The participants hopped on their dominant leg (the leg used to kick a ball for distance), contralateral knee flexed at 90-degrees and hands

non-restricted for balance (Fig 6.1). The dominant leg was utilized to standardize testing. Participants were instructed to undertake the hop as quickly as possible. A recovery of ten seconds was provided between each trial.

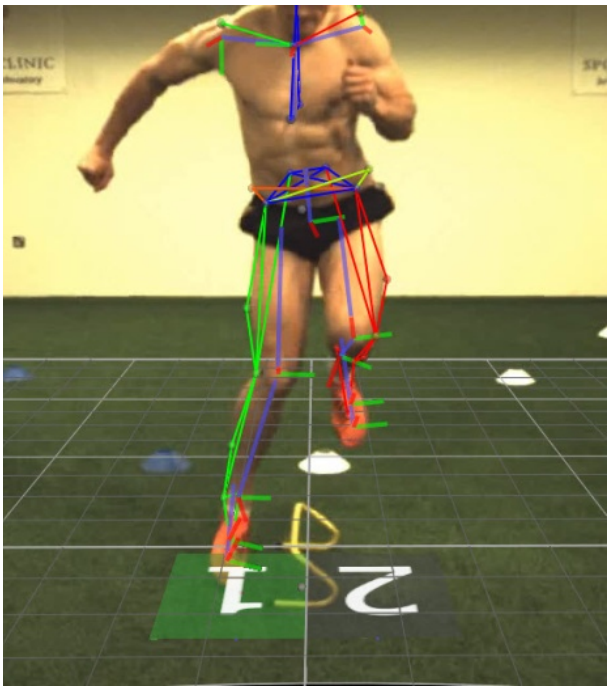

Figure 6.1: Hurdle Hop exercise depicting the initial landing of interest.

### 6.2.3 Data acquisition and analysis

Three-dimensional motion data was collected using eight infrared cameras (Vicon-Bonita B10, UK), synchronized with two 40x60cm force platforms (AMTI-BP400600, USA) to collect ground reaction forces. Reflective markers (14mm diameter) were placed at bony landmarks on the lower limbs, pelvis and trunk as per the Vicon Plug in Gait model (Marshall et al., 2016). Motion and force data was captured at 200Hz and 1000Hz, respectively, and were both filtered using a fourth order Butterworth filter with a cut-off frequency of 15Hz (Kristianslund, Krosshaug and van den Bogert, 2013). Data was statistical analysed in Excel 2007 (Microsoft, Redmond, WA, USA) and SPSS 21 (SPSS, Chicago, IL). Range of motion (ROM) at the trunk, pelvis, hip, knee and ankle, in addition to peak moments for the latter

three joints were examined during the contact phase of the first landing. Ground contact time was compared between the first and last trial using a t-test to confirm the absence of fatigue.

### 6.2.4 Methods of determining the required number of trials

**Peak Intraclass Correlation Coefficient ($ICC_p$) Method** The $ICC_p$ method assesses the inter-trial effect in relation to the inter-subject effect by utilizing a two way fixed ICC model:

$$ICC_{3,1} = \frac{MS_s - MS_e}{MS_s + (K-1) * MS_e} \tag{6.3}$$

Where $MS_s$ is the subject mean square, $MS_e$ is the mean square error and $K$ is the number of trials. To calculate the number of trials required to achieve a representative mean, each trial is added one at a time from trial number 2 to 15. At each iteration, the $ICC_{3,1}$ is calculated and the number of trials required is subsequently determined when the $ICC_{3,1}$ value reaches its peak. The implementation of this method is explained in length by (Racic, Pavic and Brownjohn, 2009).

**Sequential Analysis Methods** These methods generate a cumulative mean by adding one trial at a time. A representative mean is obtained once the cumulative mean falls within a bandwidth of precision and remains there for subsequent trials. Three forms of bandwidth were compared in this study namely:

$SD_{25}$**:** Defined by 25% of the total within-subject standard deviation

$SD_{30}$**:** Defined by 30% of the total within-subject standard deviation

$SEM_{ind}$**:** To establish this criterion bandwidth the SEM was utilized, which is defined as an estimate of the within-subject standard deviation ($SD_{ws}$). Since the SEM retains the same units as the test results and will change inversely to the reliability of the measurement at hand, it can be seen as a relative error measurement. Within this study the reliability term refers to the intra-class correlation coefficient ($ICC_{3,1}$) statistic. In addition to this, the classical SEM formula was adapted in this study so that the SD term refers to the $SD_{ws}$ for each subject, so that an individualised SEM bandwidth ($SEM_{ind}$) was utilized for each subject

.

$$SEM_{ind} = SD_{ws} * \sqrt{(1 - ICC_{3,1})} \tag{6.4}$$

In this sense the $SEM_{ind}$ will always provide a more conservative criteria than the $SD_{ws}$ since by definition the $SEM_{ind}$ can only achieve a value equal to the $SD_{ws}$ when the reliability is exactly 0 (not likely in human movement).

## 6.3 Findings

Mean ground contact time during the first trial (0.254 $\pm$ 0.03sec) was not significantly different (p= 0.825) from the final trial (0.252 $\pm$0.03sec) confirming the absence of fatigue during the tests. Peak moments and ROM findings for all variables are presented below (table 6.1). The mean number of trials required to provide a representative mean across all ROM variables ranged from 2 to 10 depending on method, joint and anatomical plane (table 6.2). Across all variables the $SEM_{ind}$ provided the least conservative method with a maximum of 6 trials. The $ICC_p$ method required the greatest number of trials with 10 trials for ankle flexion ROM. With the exception of this variable however, the SD methods required a consistently larger number of trials than the $ICC_p$ method for ROM. For peak mo-

Table 6.1: Summary of number of trials required when examining range of motion in the hurdle hop

| Joint | $ICC_p$ | | | $SEM_{ind}$ | | | $SD_{25}$ | | | $SD_{30}$ | | | Max per joint | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Sagittal | Frontal | Transverse | Sagittal | Frontal | Transverse | Sagittal | Frontal | Transverse | Sagittal | Frontal | Transverse | $ICC_p$ | $SEM_{ind}$ | $SD_{25}$ | $SD_{30}$ |
| **Ankle** | 10 | 2 | 2 | 3.17 ± 2.12 | 6.17 ± 3.41 | 3.08 ± 1.83 | 7.58 ± 3.68 | 8.0 ± 3.10 | 7.92 ± 3.63 | 7.08 ± 3.32 | 6.92 ± 3.09 | 7.0 ± 3.35 | 10 | 6 | 8 | 7 |
| **Knee** | 3 | 2 | 7 | 3.50 ± 2.11 | 4.17 ± 2.12 | 4.42 ± 1.93 | 8.33 ± 2.64 | 8.17 ± 2.52 | 6.92 ± 2.54 | 7.75 ± 3.2 | 7.75 ± 2.6 | 6.83 ± 2.59 | 7 | 4 | 8 | 8 |
| **Hip** | 2 | 7 | 2 | 4.0 ± 2.6 | 3.17 ± 2.29 | 5.0 ± 2.59 | 7.92 ± 2.54 | 7.83 ± 3.90 | 8.33 ± 2.53 | 7.50 ± 2.58 | 7.67 ± 3.50 | 7.83 ± 2.66 | 7 | 4 | 8 | 8 |
| **Pelvis** | 2 | 4 | 2 | 3.08 ± 1.38 | 3.33 ± 2.57 | 2.58 ± 1.0 | 7.33 ± 3.98 | 8.58 ± 2.75 | 8.75 ± 2.63 | 7.18 ± 3.4 | 7.92 ± 3.15 | 8.0 ± 2.49 | 4 | 3 | 9 | 8 |
| **Trunk** | 5 | 5 | 2 | 2.83 ± 1.85 | 2.67 ± 0.98 | 4.08 ± 1.38 | 7.75 ± 2.60 | 8.83 ± 2.21 | 8.58 ± 1.93 | 6.33 ± 2.57 | 8.17 ± 2.62 | 7.0 ± 2.56 | 5 | 3 | 9 | 8 |
| **Max per Plane** | 10 | 7 | 7 | 4 | 6 | 4 | 8 | 9 | 9 | 8 | 8 | 8 | | | | |

$ICC_p$ = Peak intraclass correlation coefficient method, $SEM_{ind}$ = Sequential analysis bandwidth defined by a modified version of the standard error of measurement, $SD_{25}$ & $SD_{30}$ = Sequential analysis bandwidth defined by 25 and 30% of the within subject standard deviation respectively

ments the number of trials required ranged from 3 to 12 trials (table 6.3). Overall the $SEM_{ind}$ was the least conservative method. The $ICC_p$ method demonstrated the highest peak values whilst the SD methods had generally higher values across each measurement. Across all variables the $ICC_p$ method provided the most inconsistent results; ranging from 2 - 12 trials required to achieve a representative mean, depending on the variable of interest.

Table 6.2: Summary of number of trials required when examining peak moments in the hurdle hop

| Joint | $ICC_p$ | | | $SEM_{ind}$ | | | $SD_{25}$ | | | $SD_{30}$ | | | Max per joint | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Sagittal | Frontal | Transversus | Sagittal | Frontal | Transversus | Sagittal | Frontal | Transversus | Sagittal | Frontal | Transversus | $ICC_p$ | $SEM_{ind}$ | $SD_{25}$ | $SD_{30}$ |
| **Ankle** | 3 | 3 | 3 | 5.83 ± 2.92 | 5.67 ± 2.41 | 5.0 ± 2.83 | 7.42 ± 2.75 | 9.17 ± 2.41 | 8.17 ± 2.66 | 7.17 ± 2.75 | 8.25 ± 3.02 | 7.08 ± 3.42 | 3 | 6 | 9 | 8 |
| **Knee** | 6 | 11 | 4 | 4.92 ± 2.61 | 6.25 ± 2.26 | 5.58 ± 3.29 | 9.08 ± 3.15 | 8.58 ± 2.87 | 8.75 ± 2.26 | 8.83 ± 3.27 | 7.58 ± 2.64 | 6.92 ± 3.00 | 11 | 6 | 9 | 9 |
| **Hip** | 4 | 6 | 12 | 4.33 ± 2.64 | 6.42 ± 3.29 | 2.58 ± 0.67 | 7.92 ± 2.91 | 8.33 ± 2.64 | 8.42 ± 2.81 | 7.42 ± 2.87 | 7.0 ± 3.13 | 7.33 ± 2.10 | 12 | 6 | 8 | 7 |
| **Max per plane** | 6 | 11 | 12 | 6 | 6 | 6 | 9 | 9 | 9 | 9 | 8 | 7 | | | | |

$ICC_p$ = Peak intraclass correlation coefficient method, $SEM_{ind}$ = Sequential analysis bandwidth defined by a modified version of the standard error of measurement, $SD_{25}$ & $SD_{30}$ = Sequential analysis bandwidth defined by 25 and 30% of the within subject standard deviation respectively

Table 6.3: Mean and standard deviation (SD) for range of motion and peak moments in the hurdle hop.

| | Range of Motion | | Peak Moments Nm/Kg | |
|---|---|---|---|---|
| **Variable** | **Mean** | **SD** | **Mean** | **SD** |
| **Ankle Flex/Extension** | 40.21 | 3.61 | 3.46 | 0.81 |
| **Ankle Ab/Adduction** | 2.12 | 1.07 | 0.24 | 0.13 |
| **Ankle Int/Ext Rotation** | 19.55 | 4.80 | 0.55 | 0.18 |
| **Knee Flex/Extension** | 30.49 | 7.04 | 3.22 | 0.68 |
| **Knee Ab/Adduction** | 7.72 | 2.49 | 1.73 | 0.55 |
| **Knee Int/Ext Rotation** | 17.23 | 4.15 | 0.31 | 0.14 |
| **Hip Flex/Extension** | 15.79 | 5.00 | 1.86 | 0.58 |
| **Hip Ab/Adduction** | 10.74 | 2.78 | 1.78 | 0.57 |
| **Hip Int/Ext Rotation** | 11.13 | 4.86 | 0.13 | 0.04 |
| **Pelvis Ant/Posterior tilt** | 6.33 | 1.69 | - | - |
| **Pelvis Ipsi/contralateral tilt** | 7.22 | 2.47 | - | - |
| **Thorax Ant/posterior tilt** | 6.54 | 2.08 | - | - |
| **Thorax Ipsi/contra Flexion** | 6.08 | 1.62 | - | - |
| **Thorax Flex/Extension** | 4.16 | 0.68 | - | - |
| **Thorax Ipsi/Contra Rotation** | 8.22 | 3.72 | - | - |

## 6.4 Discussion

The number of trials obtained during an experiment can affect how representative the mean will be (Hamill and McNiven, 1990; Racic, Pavic and Brownjohn, 2009). Despite this, few authors have presented empirical evidence for the number of trials utilized (e.g. (Moran and Wallace, 2007; Amiri-Khorasani, Osman and Yusof, 2010; Villeger et al., 2014)). The purpose of this study was to compare four different methods to determine the number of trials required to obtain a stable representation of the mean during a hurdle hop movement. In accordance with previous investigations (James et al., 2007; Racic, Pavic and Brownjohn, 2009) the present study identified that different methods produce dissimilar results. Predictably the two methods that produced the most similar results were $SD_{25}$ and $SD_{30}$ bandwidths. In general the $SD_{25}$ bandwidth was the most conservative, whilst the $SEM_{ind}$ required the least number of trials. It should be noted that only male subjects were utilized in this study to ensure a homogeneous group. As such generalization of this study's results to different populations should be made with caution. Furthermore since no standardised statistical power tests are available for the analysis utilised in this study, 15 subjects were selected similar to previous research (James et al., 2007; Racic, Pavic and Brownjohn, 2009; Taylor et al., 2015). There are potential limitations to the methods presented in this study. The sequential method as proposed in the past uses an arbitrary choice of a cut-off bandwidth, which is open to the potential of bias. This is clearly evident in the paper by Rodano and Squadrone (Rodano and Squadrone, 2002), in which the authors choose by trial and error a criterion bandwidth that provides a stable and consistent mean. The $ICC_p$ method on the other hand is subject to potentially falsely identifying two trials as stable should they, by chance, be highly correlated. Indeed in this study many variables demonstrated a requirement of two trials when utilizing the $ICC_p$ method. Furthermore, since the $ICC_{3,1}$ statistic

is a ratio of the differences between subjects to the variability in the data, the ICC is highly influenced by between-subject variability which when high will improve the ICC score obtained(Weir, 2005). Whilst the proposed $SEM_{ind}$ method is also subject to this limitation since it uses the $ICC_{3,1}$, it is offset to some extent by the $SD_{ws}$ in its calculation making it modestly affected by between subjects variability. Unlike the $ICC_p$ method, which is based on the entire group, the $SEM_{ind}$ represents a relative error measurement that is individualised to each participant. The limitations of this study include the use of 15 trials. Previous research (Taylor et al., 2015) has demonstrated that the total number of trials can affect the results obtained from a sequential analysis. 15 trials were utilized in this study as it was felt to be the largest number of trials that could be collected within the available time. Future research should ascertain if these results hold true for different trial numbers, across different populations and using a continuous waveform approach rather than discrete points.

## 6.5 Conclusion

In summary, several methods were compared within this research. Interestingly different methods can produce strikingly different results for the same task and variable. The $SEM_{ind}$ method produced the least conservative results and was deemed the most practical in human motion capture where fatigue may be an issue. Whilst not free of limitations, the $SEM_{ind}$ method presented the least limitations. A consensus on which method to adopt should be reached, however for reasons stated above we recommend the novel $SEM_{ind}$ method presented in this research. Regardless of the method chosen, researchers should begin providing empirical evidence for the number of trials used.

### 6.5.1 Relevance of this Chapter's findings to this thesis

The aim of this Chapter was to determine the number of trials required to obtain a representative mean of the hurdle hop task. This was conducted in parallel to data capture for Chapter 3, 4 and 5. The results from this study indicate that a total of 6 trials were required to obtain a representative movement pattern during a lateral hurdle hop test. It is acknowledged that utilising a smaller number of trials (as conducted in this PhD thesis) may have resulted in unreliable findings for Chapter 3. It is unknown how the findings from this study would affect the results from Chapter 4 and 5 which examined stiffness and complexity respectively.

### 6.5.2 Link between Chapter 6 and Chapter 7

The experimental investigations from Chapter 3 to 5, have utilised a hurdle hop test to investigate the biomechanics of AGP. When investigating the biomechanics associated with an injury however, it is important that biomechanical screening replicate the demands of the actions associated with the injury (Marshall et al., 2016). The running cut task may therefore provide a more ecologically valid examination of AGP.

Furthermore, when examining the biomechanical risk factors for AGP, Chapter 3 to 5 utilised a single group study design. Such an approach assumes that the single group examined is suitably homogeneous in nature. However, recently our research group have demonstrated the presence of biomechanical sub-clusters of AGP patients during a running cut task (Franklyn-Miller et al., 2017). Therefore the aim of Chapter 7 was to explore the changes in a running cut test that occur from pre- to post- rehabilitation within the sub clusters identified by Franklyn-Miller et al., (2017).

# Chapter 7

# Cluster specific biomechanical changes post- rehabilitation and their association with change in an outcome measure

## 7.1 Introduction

Athletic groin pain (AGP) is a common injury in sports involving agility-based movements (Werner et al., 2009; Orchard, Seward and Orchard, 2013; Thorborg et al., 2017) and can be typically characterised by an insidious onset of chronic pain to the lower abdominal and/or pubic region (Falvey, King and Kinsella, 2015). It is thought that this chronic pain is caused by a repetitive overload to the structures surrounding and attaching into the anterior pubic symphysis. As such there has been growing interest in the association between AGP and movement biomechanics, especially during agility tasks (Edwards, Brooke and Cook, 2017; Franklyn-Miller et al., 2017; King et al., 2018).

In the absence of prospective research, the biomechanical comparisons of injured versus uninjured and/or injured versus rehabilitated are commonly utilized approaches to identify biomechanical variables associated with AGP (Edwards, Brooke and Cook, 2017; Gore et al., 2018; King et al., 2018). However a potentially more informative approach is to identify the relationship between the change in biomechanics and the change in return to play (RTP) status, as the biomechanical factors that change with rehabilitation that are deterministic of a successful rehabilitation should be related to change in RTP status. This is in line with the probabilistic approach to causation (Burr, 2003; Marshall and Moran, 2015) which defines causation in terms of a cause preceding and increasing the probability of the effect. Despite this, examinations of this nature have not been conducted in AGP research. A challenge to examining the association between biomechanics and RTP status is that the RTP metric is binary in nature (successful vs. unsuccessful) and hence provides somewhat limited information regarding the quality of rehabilitation. In the absence of a RTP scale, it is important to examine an alternative outcome scale of 'pain and function' that is deemed to be representative of the RTP status (e.g. the HAGOS Function in Sport and Recreation subscale).

Research typically examines the biomechanical risk factors for an injury using a single-group study design. Such an approach assumes that the single group examined is suitably homogeneous in nature, that is, there is no underlying participant characteristic that may confound the study's findings. A potential limitation to this approach is the requirement for a priori knowledge of these potentially confounding factors. An alternative method is to utilise a statistical sub-grouping technique known as clustering to identify similar subgroups within a larger sample/population (Rein et al., 2010). For example, Phinyomark et al. (2015) identified two distinct running gait patterns in a group of healthy runners independent of age, height, weight, and running speed. When these two groups were separately compared to a large cohort of runners experiencing patellofemoral

pain, two different risk factors were identified in relation to knee abduction angle. This highlights the potential importance of using clustering techniques to identify movement sub clusters within a larger apparently homogeneous sample.

Recently our research group have demonstrated the presence of three distinct biomechanical cluster subgroups of AGP patients during a running cut task (Franklyn-Miller et al., 2017).The motivation and hypothesis was that the identified subgroups would be related to the specific anatomical diagnosis. Interestingly, the authors found no relationship between anatomical diagnosis and the movement subgroups, suggesting that rehabilitation should be specifically tailored for each subgroup.

To date however, no research has demonstrated that these clusters respond differently to a criterion-based rehabilitation programme. The aim of this study therefore, was to quantify the relationship between the change in biomechanics and the change in 'pain and function' following rehabilitation within each cluster. It was hypothesised that the three clusters would demonstrate some distinct biomechanical changes in response to a criterion-based rehabilitation programme.

## 7.2 Methods

Three hundred and twenty two male subjects with AGP who presented to the sports medicine department of the Sports Surgery Clinic, Dublin from January 2013 to May 2015 were assessed for eligibility in this study. Subjects presenting with exercise-induced pain in the groin area (Falvey, Franklyn-Miller and McCrory, 2009), regardless of the painful anatomical structure involved, were assessed. All AGP subjects were reviewed by a Consultant Physician in Sport and Exercise Medicine as per Falvey, King and Kinsella, (2015). Exclusion criteria included the presence of hip joint arthrosis [grade 3 or higher on MRI (Li et al., 1988)], those who did not intend to return to pre-injury activity levels, those with symptoms less

than four weeks, those with underlying medical conditions such as inflammatory arthropathy or infection, those who did not consent to be a part of the study and those who did not successfully complete the exercise rehabilitation program as outlined in King et al., (2018). Of the 322 subjects, 104 met the full study criteria. Due to missing and corrupt data, 86 subjects were subsequently examined in this investigation. The Sports Surgery Clinic Hospital Ethics Committee approved the study (Ref 25EF011) and all subjects signed informed consent. The study was registered at Clinicaltrials.gov (NCT02437942).

### 7.2.1 Clinical intervention

All subjects undertook a criterion-based rehabilitation programme with three levels of progression, which focused on control of the hip, pelvis and trunk during dynamic loading tasks. Level 1 addressed inter-segmental control and strength, level 2 involved linear running mechanics and increasing linear running load tolerance, and level 3 addressed multidirectional mechanics and the transition back to high intensity sprinting. Recovery was defined as when the patient progressed through level 3 of the rehabilitation program and returned to pain free participation in sport. Advancement through the levels of progression was individualised based on each subject achieving key goals for progression. A full description of this intervention has been published previously (King et al., 2018) in accordance with the TIDieR (template for intervention description and replication) checklist and guide (Hoffmann et al., 2014) and is presented in Appendix C.

### 7.2.2 Biomechanical Protocol

This study involved each AGP subject attending the lab on two occasions (pre- and post- rehabilitation). Prior to the experimental testing, the subjects completed a standardised dynamic warm-up (including five body weight squats and five

submaximal countermovement jumps). The experimental testing involved the participants completing three maximal effort planned 110° cuts on artificial turf (Figure 7.1). Each subject undertook two submaximal practice trials of the cut before test trials were captured on both the right and left leg. The order of testing was randomised and a 1 min recovery was taken between trials.

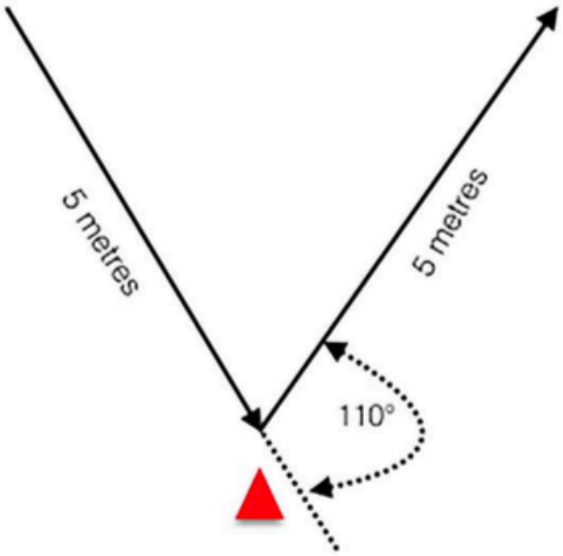


Figure 7.1: Illustration of the 110° cut task

### 7.2.3 Data Acquisition

Twenty-eight reflective markers (14mm diameter) were placed at bony landmarks on the lower limbs, pelvis and trunk as per the Vicon Plug in Gait model (Vicon Motion Systems, Oxford, UK). Lower extremity and trunk kinematics and kinetics were captured during each trial. Three dimensional marker trajectories were tracked using 8 infrared cameras (Vicon - Bonita B10, UK), and with two synchronized 40x60cm force platforms (AMTI - BP400600, USA) collecting ground reaction force data. Motion and force data were captured at a sampling frequency of 200Hz and 1000Hz, respectively. The Copenhagen Hip And Groin Outcome Score (HAGOS) was collected pre and post rehabilitation as a validated measure of pain and function in athletes (Thorborg et al. 2011).

### 7.2.4 Data Processing

Both marker and force data were filtered using a fourth order Butterworth filter with a cut-off frequency of 15Hz (Kristianslund, Krosshaug and van den Bogert, 2013). Kinematic and kinetic calculations were performed in Nexus software (Vicon Motion Systems, Oxford, UK) and were subsequently exported to Matlab for statistical analysis (R2015b, MathWorks Inc., USA). Kinematic and kinetic variables were defined as per the standard Vicon Plug in Gait model. Additional variables examined included joint angular velocity, joint work, the distance between the centre of mass and the centre of pressure and the thorax relative to pelvis angle (King et al., 2018).A summary of the biomechanical variables examined in this chapter is presented in Table 7.1. No knee alignment device was utilised in this study. While some concern has been raised with regards to the repeatability and accuracy of the knee joint axes in the plug in gait model (Baudet et al., 2014), this model is widely used to explore the mechanics of various injuries during dynamic tasks (Russell et al., 2006; Bencke et al., 2013; Ali, Robertson and Rouhi, 2014; Gribbin et al., 2016; Schreurs, Benjaminse and Lemmink, 2017) and has been shown by our research group to have high repeatability in the running cut (ICC = 0.90) when utilising our lab's methodology (Marshall et al., 2015).

The leg examined was the painful side, or in the case of bilateral pain the side with the most pain, as determined via clinical palpation. Where equal pain was found on both sides, the leg examined was determined randomly. Where no palpation pain was identified, the side of MRI confirmed pathology was used. Again, where both sides were affected, the leg examined was selected randomly.

The data was examined during ground contact defined by a 5N threshold of the vertical ground reaction force. The data was normalised from 0% to 100% of the ground contact using 101 time nodes and landmark registered to the start of the concentric phase using dynamic time warping (Ramsay and Silverman,

Table 7.1: Summary of Biomechanical variables examined

| Variable | Description of Calculation |
|---|---|
| **COM height** | Vertical position of the COM. |
| **COM power** ‡ | (Ground reaction force * COM displacement)/time. |
| **Ground contact time** ‡ | Time from initial contact to toe off defined as when the vertical ground reaction force passes a 5N threshold. |
| **Ankle, Knee and Hip joint angle** | Relevant angle between two adjacent segments described with respect to the proximal coordinate system. |
| **Thorax and Pelvis angle** | Absolute angle relative to global coordinate system. |
| **Thorax relative to Pelvis angle** ‡ | Relative angle between the Thorax and Pelvis described with respect to the proximal coordinate system. |
| **Joint/ Segment velocity** ‡ | Derivative of joint/ segment angle. |
| **Joint/ Segment moments** | Calculated using inverse dynamics and presented in the local co-ordinate frame of the distal segment. |
| **Joint/ Segment powers** | Moment * angular velocity. |
| **Joint/ Segment work** ‡ | Moment * angular displacement. |
| **Resultant impulse** ‡ | $\sqrt{[(Force_x * time)^2 + (Force_y * time)^2 + [(Force_z * time)^2]}$ |
| **COM to Centre of pressure** ‡ | Eudiclean distance between COM and centre of pressure |

COM = Centre of mass; ‡ = Manually calculated variable.

2005). The AGP patients were clustered into three clusters as previously described using analysis of characterising phases and a hierarchical clustering approach (Franklyn-Miller et al., 2017). The hierarchical clustering approach utilized a gap statistic to decide the number of clusters. Subject scores were generated using analysis of characterising phases and normalized into a correlation matrix, which were then used as an input for gap statistic and the hierarchical clustering approach.

To identify if there was an overall change in the performance outcome of the cutting task, the resultant impulse was calculated in both the deceleration and concentric phase of the examined pivot step. This metric was chosen as impulse generated when in contact with the ground directly determines the change in velocity (as per the impulse-momentum relationship).

### 7.2.5 Data Analysis

To quantify the relationship between the change in biomechanics and the change in pain and function following rehabilitation within each of the three clusters, a three-step process (feature generation and selection, model creation and model testing) was utilised to find the model that best predicts the change in the HAGOS Function in Sport and Recreation subscale (HAGOS-FSR). This subsection of the HAGOS questionnaire was selected as it was felt it best represented the symptoms of AGP.

**Step 1: Feature generation and selection** The first step was to identify key phases of variation within the biomechanical waveforms using analysis of characterising phases (Richter, O'Connor, et al., 2014a). Using a smallest worthwhile Cohen's d effect size change of 0.2 (Cohen, 1988) and 95% confidence limits, a key phase was retained only when a clear substantial change was identified from pre- to post-rehabilitation (Batterham and Hopkins, 2006). To maximize the ability of each feature to explain the change in HAGOS-FSR, the retained phases were then dynamically extended until they reached a point where the absolute bivariate correlation between the mean change in each phase and change in the HAGOS-FSR would decrease and/or the phase would no longer demonstrate a clear substantial change from pre- to post- rehabilitation. Any phase less than 5% in length was then discarded to maintain only functionally meaningful features (Richter et al., 2017). To prevent over fitting and to improve the generalisability of our findings, elastic net regularization was utilised (Zou and Hastie, 2005). To select the two tuning parameters ($\alpha$ and $\lambda$) required for the elastic net, a five fold cross-validation approach was utilised similar to that previously proposed (Zou and Hastie, 2005)[1]. The $\alpha$ and $\lambda$ values that produced the lowest cross-validated

[1]Twenty values of $\alpha$ ranging from 0 to 1 were tested and for each $\alpha$ value, 100 $\lambda$ values were generated.

root mean square error (cvRMSE) was utilised in the model and the biomechanical features selected by the elastic net model were recorded. This was conducted on a random 70% subset of the data one hundred times.

**Step 2: Model generation** The 15% most commonly extracted variables from step 1 were then retained and using all the data available, the final model was selected using both a test-all subsets approach and the cross validated elastic net method (as described in step 1) for each cluster. The final model chosen was the model which minimized the cross validated RMSE.

**Step 3: Model testing** To provide confidence in the cluster specific models generated, it was important that the generated models were unique to each cluster and provided greater prediction ability in comparison to a model generated at a whole group level (all clusters combined). To determine if the generated models were unique to each cluster, the features identified within each cluster were shuffled amongst the other clusters to predict change in HAGOS-FSR. For example, after the final model features were identified in cluster 1, the same features were then extracted from the data of cluster 2 and cluster 3 and used to predict change in HAGOS-FSR in the respective clusters (Figure 7.2). To determine if the models generated at a cluster level provided better prediction ability in comparison to a model generated at a whole group level, steps 1 - 2 were also repeated at a whole group level and compared to the cluster specific models.

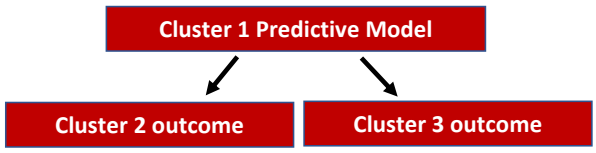


Figure 7.2: Illustration of the between cluster model test. Each Cluster's predictive model is applied to each Cluster's Data to predict the change in injury status.

Post rehabilitation, the AGP subjects were classified using a correlation approach outlined in Richter et al. (2017). This was conducted to understand how the

change in biomechanics from pre- to post-rehabilitation affected cluster membership. This may be important to enhance our understanding of how rehabilitation affects the mechanics of cutting in AGP patients. Clinical and demographic measures were also compared between the three clusters using an analysis of variance. Post hoc examinations, where required, were conducted using a Tukey's honest significant difference test. The alpha level was set at 0.05 and all statistical analysis was undertaken in Matlab 2015b (Mathworks, USA).

## 7.3 Findings

There were no significant differences between the three clusters in anthropometric measures or weeks with pain on presentation, however there was a significant difference in the age of the clusters with cluster 3 significantly older than cluster 2 (Table 7.2).

All three clusters improved significantly in all HAGOS scores from pre- to post-rehabilitation. There were no significant differences between the clusters in HAGOS score pre-rehabilitation or in the change of HAGOS score from pre- to post-rehabilitation (Table 7.3).

The mean amount of time in days until return to play (Cluster 1: 75.8±42.9, Cluster 2: 75.7±27.1, Cluster 3: 70.1±26) was also not significantly different between the three clusters (p=0.77). Resultant concentric impulse, which represents

Table 7.2: Descriptive metrics of the three clusters

| | Age (yrs.) | Height (m) | Weight (kg) | Weeks with Pain |
|---|---|---|---|---|
| **Cluster1** | 25.1±5.0 | 1.79 ± 0.06 | 79.31 ± 8.8 | 55.35 ± 67.4 |
| **Cluster2** | 22.7±4.8 | 1.78 ± 0.07 | 77.72 ± 9.7 | 63.84 ± 86.6 |
| **Cluster3** | 27.2±3.7 | 1.8 ± 0.05 | 81.11 ± 9.1 | 39.23 ± 36.1 |
| **One Way ANOVA (p)** | 0.04 | 0.55 | 0.37 | 0.39 |
| **Post hoc** | C3 > C2 (p = 0.04) | - | - | - |

yrs. - years, m - meters, kg - kilograms

Table 7.3: HAGOS findings for the three clusters both pre- and post-rehabilitation

| HAGOS | Cluster 1 | | Cluster 2 | | Cluster 3 | |
|---|---|---|---|---|---|---|
| | Pre | Post | Pre | Post | Pre | Post |
| **Pain** | 72.8 ± 17.2 | 83.8 ± 13.8 | 75.7 ± 11.8 | 85.8 ± 13.5 | 75.3 ± 14.9 | 86.0 ± 12.6 |
| **Symptoms** | 62.6 ± 18.4 | 87.2 ± 11.8 | 62.0 ± 16.7 | 88.3 ± 11.0 | 63.7 ± 17.1 | 89.2 ± 9.5 |
| **ADL** | 72.3 ± 21.7 | 90.3 ± 13.0 | 73.6 ± 14.6 | 93.8 ± 11.2 | 77.0 ± 18.2 | 92.0 ± 13.5 |
| **Sports** | 53.3 ± 17.2 | 84.2 ± 13.6 | 49.4 ± 17.3 | 85.1 ± 13.6 | 53.7 ± 18.7 | 86.0 ± 13.3 |
| **PA** | 33.1 ± 33.0 | 50.6 ± 31.8 | 42.5 ± 41.9 | 59.5 ± 33.9 | 25.5 ± 27.0 | 60.3 ± 31.9 |
| **QOL** | 34.3 ± 14.1 | 59.3 ± 19.6 | 36.6 ± 14.0 | 57.4 ± 23.0 | 34.3 ± 14.9 | 65.2 ± 23.4 |

ADL - Activities of Daily Living; Sports - Sport and Recreational Activities; PA - Participation in Physical Activity; QOL - Quality of Living

the change in speed during the concentric phase, increased significantly in all three clusters from pre- to post-rehabilitation ($p<0.05$).

In relation to the main aim of this study, the ability of biomechanics to predict change in HAGOS-FSR within each cluster is detailed in Table 7.4 [2]. The features selected in Cluster 1 had the least predictive power of the three clusters. However, the model, which contained nine variables, could still explain 77% of the variation in the change in HAGOS-FSR and had a cross-validated predicted $R^2$ of 0.74 and a cross-validated RMSE of 7.36. For cluster 2, the features selected had the highest predictive power, and a model containing seven variables explained 93% of the variance in the change of HAGOS-FSR. This model also performed the best on the cross-validated data with a mean predicted $R^2$ of 0.91 and a cross-validated RMSE of 4.31. Cluster 3, had a model containing eight variables that could explain 92% of the variation in change of HAGOS-FSR. This model also preformed well on cross-validated data with a mean predicted $R^2$ of 0.85 and a RMSE of 4.80. The key changes required for each cluster group is presented in Figure 7.3 while bivariate plots of the model features are presented in Figures 7.4 - 7.6.

The ability of change in biomechanics to predict change in HAGOS-FSR at a whole group level was lower than when examining changes at each cluster level. The whole group model contained nine variables and could explain only 48% of

[2]Generally, the higher the $R^2$ value and lower the RMSE value is, then the better the model is able to predict the dependent variable. Further, the cross-validated values provide an indication for how well the models will perform on new data.

the variance in the change of HAGOS-FSR and had a cross validated predicted $R^2$ of 0.46 and a cross-validated RMSE of 11.99.

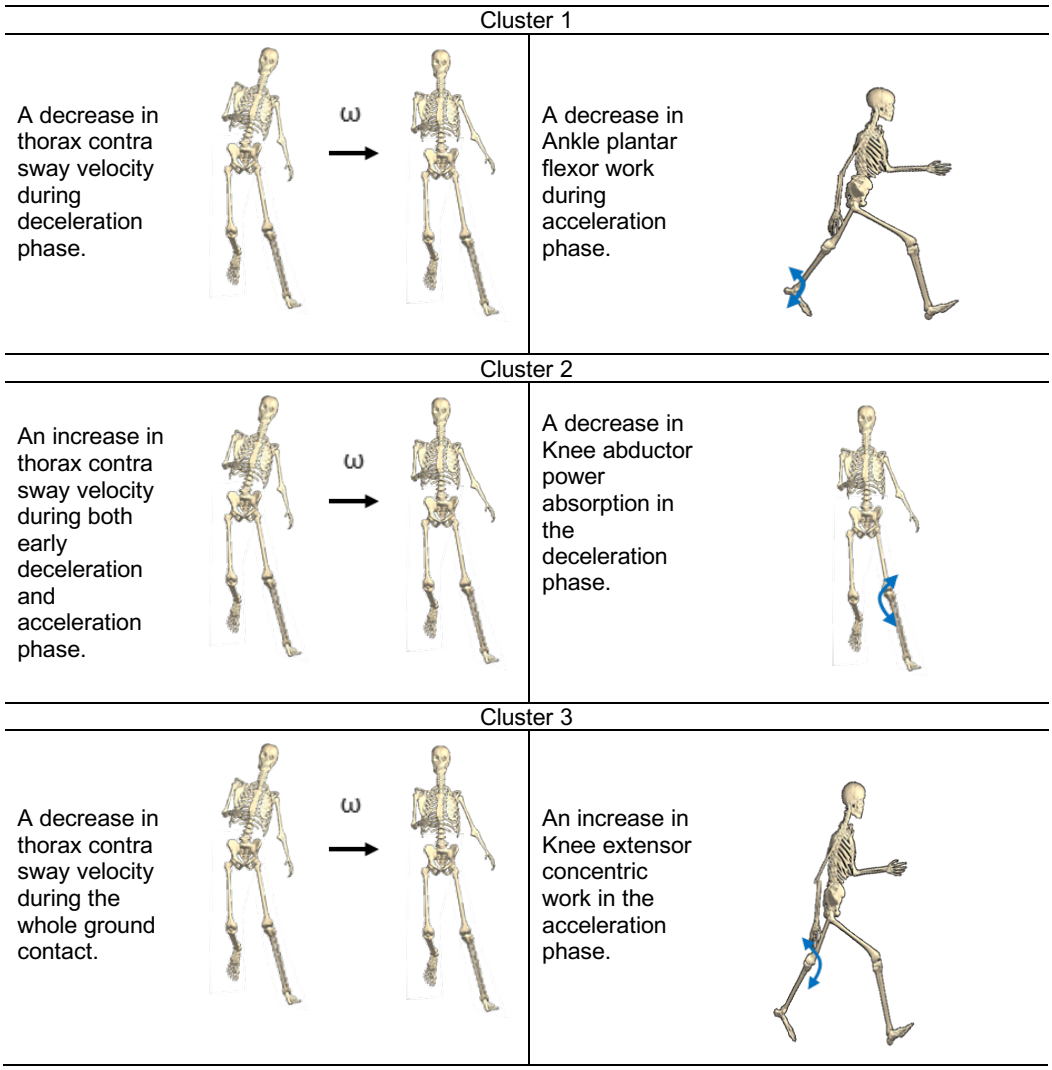


Figure 7.3: Key changes required to improve injury status for each Cluster group. $\omega$ = Angular Velocity

Table 7.4: Retained Features to explain the change in HAGOS-FSR

| **Cluster 1** ($_{CV}$RMSE = 7.36, predicted $_{CV}R^2$ = 0.74, $R^2$ = 0.77) | | |
|---|---|---|
| **Mean change in model variables** | **Beta** | **Bivariate r** |
| **Increased** | | |
| Thorax contralateral sway velocity (28-37%) | -0.30 | -0.50 |
| Ankle plantar flexor concentric work (64-100%) | -0.25 | -0.56 |
| Thorax to Pelvis contralateral sway velocity (25-44%) | 0.24 | 0.56 |
| Ankle external rotator eccentric work (33-38%) | -0.22 | -0.35 |
| Hip internal rotator moment (40-55%) | -0.19 | -0.25 |
| Pelvis anterior tilt velocity (64-73%) | -0.15 | -0.41 |
| Knee valgus power generation (41-48%) | -0.12 | -0.29 |
| **Decreased** | | |
| Knee internal rotation (0-6%) | -0.23 | -0.29 |
| Hip flexor power absorption (90-100%) | -0.21 | -0.28 |
| **Cluster 2** ($_{CV}$RMSE = 4.31, predicted $_{CV}R^2$ = 0.91, $R^2$ = 0.93) | | |
| **Mean change in model variables** | **Beta** | **Bivariate r** |
| **Increased** | | |
| Knee varus power absorption (20-26%) | 0.58 | 0.63 |
| Thorax to Pelvis contralateral sway velocity (75-80%) | -0.38 | -0.32 |
| Thorax to Pelvis contralateral sway velocity (0-23%) | -0.36 | -0.40 |
| Pelvis ipsilateral sway velocity (20-48%) | 0.28 | 0.26 |
| Ankle plantar flexion (0-6%) | 0.27 | 0.58 |
| Knee extensor power generation (45-61%) | 0.09 | -0.49 |
| **Decreased** | | |
| Ankle evertor power generation (9-27%) | -0.28 | -0.40 |

Table 7.4: Retained Features to explain the change in HAGOS-FSR (cont.)

**Cluster 3** ($_{CV}$RMSE = 4.80, predicted $_{CV}R^2$ = 0.86, $R^2$ = 0.92)

| Mean change of model variables | Beta | Bivariate r |
|---|---|---|
| **Increased** | | |
| Knee extensor concentric work (61-82%) | 0.74 | 0.53 |
| Thorax contralateral sway velocity (0-100%) | -0.70 | -0.63 |
| Ankle plantar flexor concentric work (51-68%) | 0.66 | -0.57 |
| Thorax to Pelvis extension velocity (38-73%) | -0.47 | -0.61 |
| Knee external rotator eccentric work (48-63%) | 0.28 | -0.50 |
| Hip abduction velocity (58-67%) | 0.19 | -0.44 |
| **Decreased** | | |
| Hip extensor moment (18-23%) | 0.61 | 0.34 |
| Pelvis anterior tilt velocity (39-50%) | -0.47 | -0.41 |

cvRMSE - cross validated root mean square error, $cvR^2$ - cross validated predicted coefficient of determination, $R^2$ - coefficient of determination, Beta - Standardised Beta coefficients, Bivariate r - Bivariate correlations. The absolute bivariate correlation indicates the strength of a single variable's association with HAGOS-FSR in isolation, while the standardised Beta coefficients indicates the strength of a single variable's association with HAGOS-FSR when other features are included in a model.

The ability of the identified biomechanical features in each cluster to explain variance in the other clusters (see methods: Model testing) was low with $R^2$ values ranging from 0.14 to 0.42 (Table 7.5). Post- rehabilitation, 48% of the all subjects changed cluster membership (Table 7.6).

Table 7.5: Variance explained ($R^2$) by the identified features within each cluster when applied to the other cluster groups

| | Cluster 1 data | Cluster 2 data | Cluster 3 data |
|---|---|---|---|
| **Cluster 1 variables** | 0.77 | 0.37 | 0.42 |
| **Cluster 2 variables** | 0.26 | 0.93 | 0.15 |
| **Cluster 3 variables** | 0.14 | 0.39 | 0.92 |

Note: The shaded cells reflect when a cluster's features are applied to its own cluster.

Table 7.6: Cluster membership post-rehabilitation

| | | Post-rehab Cluster | | |
|---|---|---|---|---|
| **Pre-rehab Cluster** | **Total pre-rehab** | **Cluster 1** | **Cluster 2** | **Cluster 3** |
| **Cluster 1** | 46 | 33 (72%) | 5 (11%) | 8 (17%) |
| **Cluster 2** | 31 | 10 (32%) | 7 (23%) | 14 (45%) |
| **Cluster 3** | 27 | 10 (37%) | 3 (11%) | 14 (52%) |
| **Total post-rehab** | | 53 | 15 | 36 |

Rehab = Rehabilitation. Note: The shaded cells reflect the number of subjects that did not change cluster post rehabilitation.

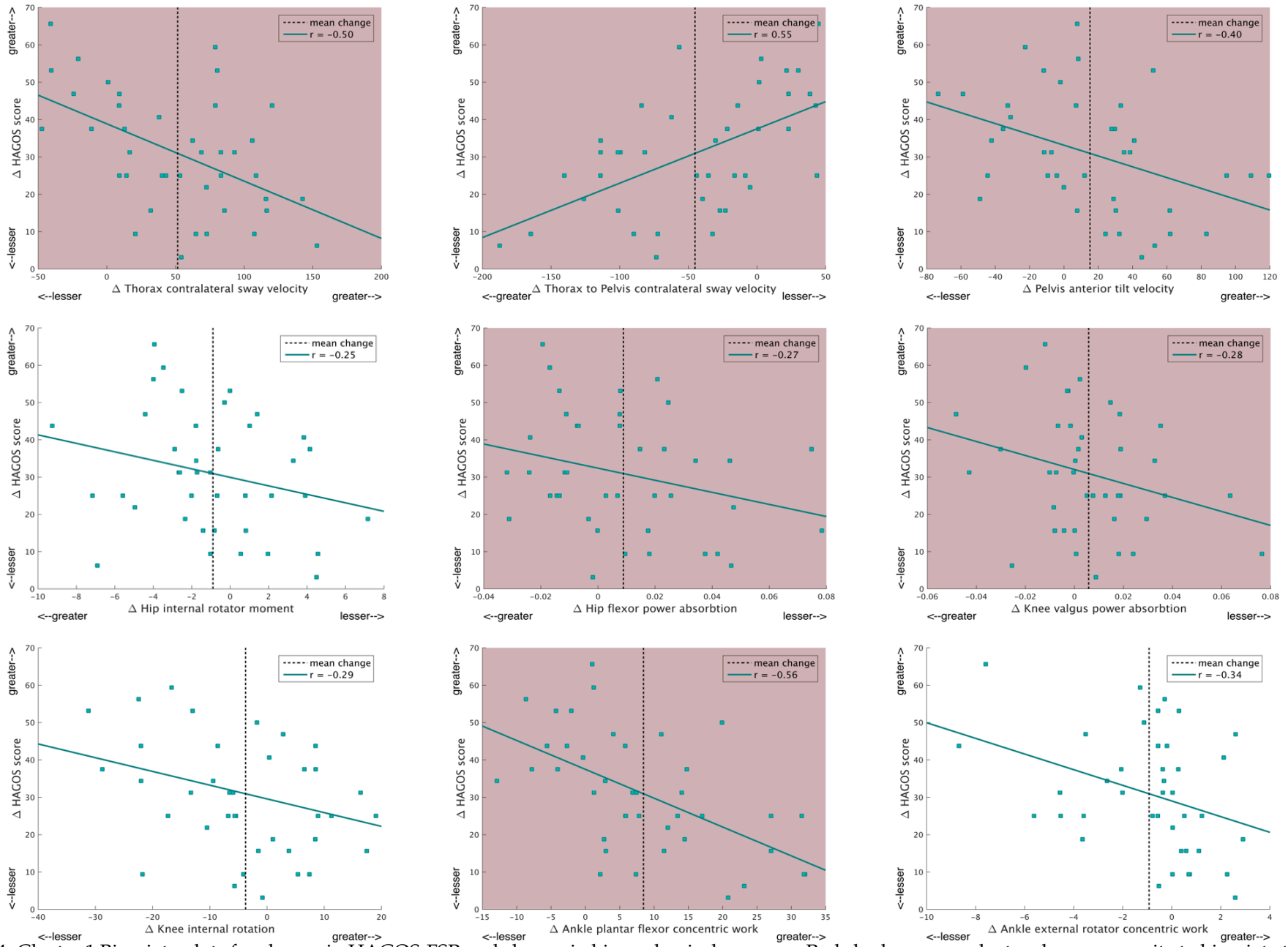


Figure 7.4: Cluster 1 Bivariate plots for change in HAGOS-FSR and change in biomechanical measure. Red shade = mean cluster change opposite to bivariate trend, Δ = change

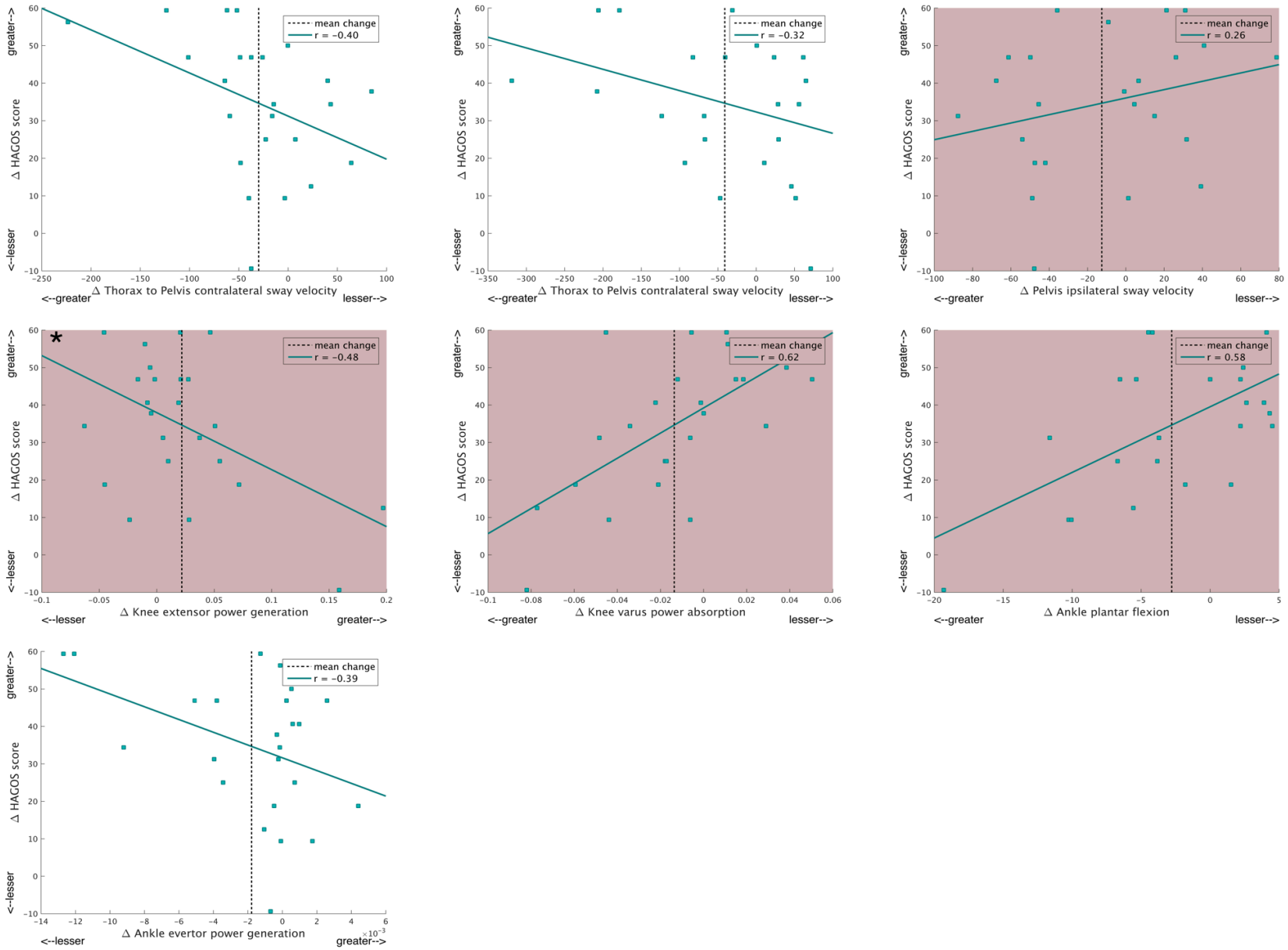


Figure 7.5: Cluster 2 Bivariate plots for change in HAGOS-FSR and change in biomechanical measure. Red shade = mean cluster change opposite to bivariate trend, Δ = change

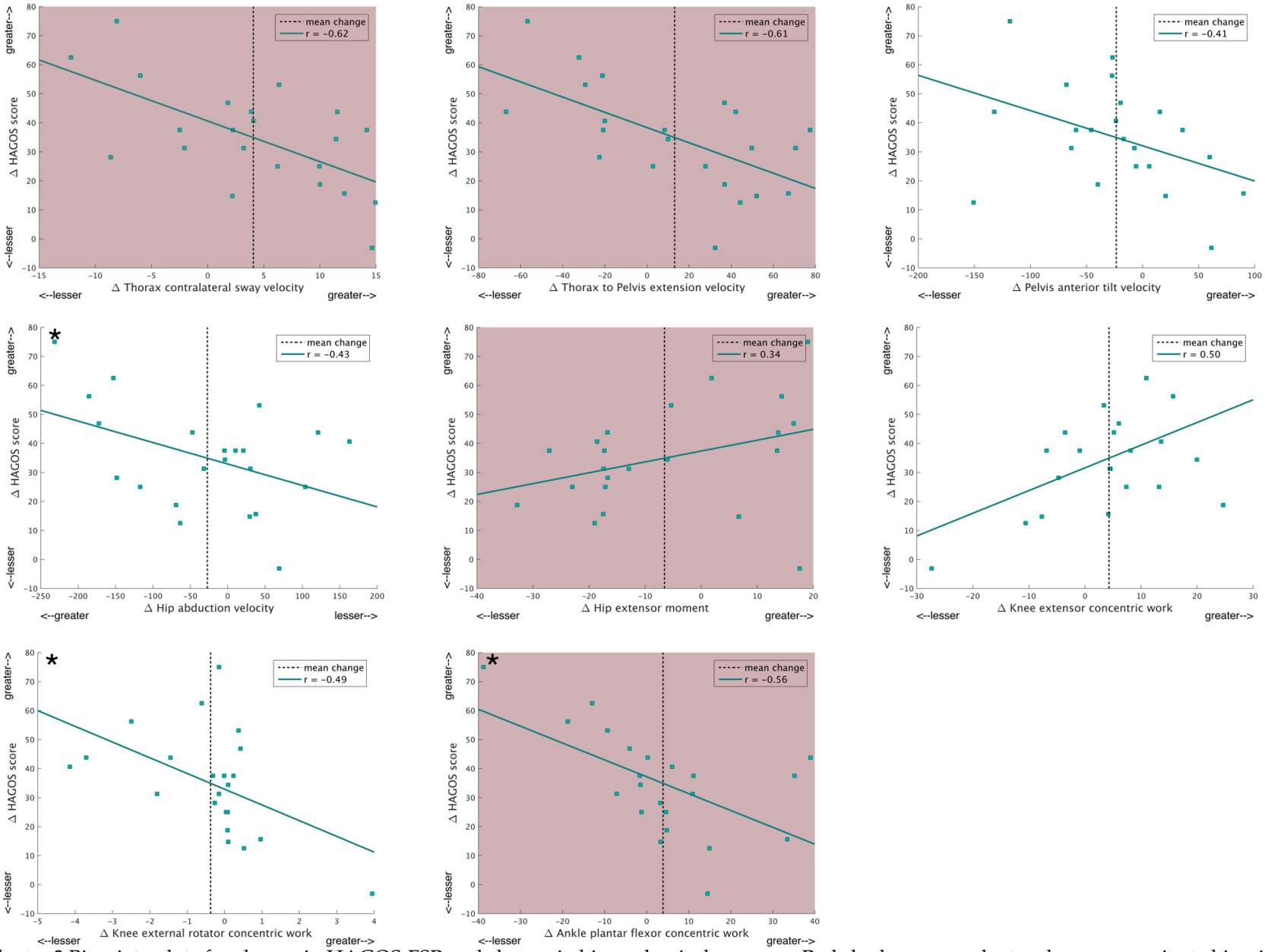


Figure 7.6: Cluster 3 Bivariate plots for change in HAGOS-FSR and change in biomechanical measure. Red shade = mean cluster change opposite to bivariate trend, Δ = change

## 7.4 Discussion

This is the first study to investigate the association between change in biomechanics from pre- to post-rehabilitation and the change in an outcome measure [HAGOS Function in sport and recreation subscale (HAGOS-FSR)] representing pain and function during dynamic tasks in AGP patients. Previous research identified the presence of three AGP cluster groups during a 110° running cut and subsequently suggested that rehabilitation should be tailored for these clusters (Franklyn-Miller et al., 2017). However, the authors provided no evidence that the cluster biomechanics would respond differently to rehabilitation. The findings from our study support our hypothesis that the three movement clusters would respond differently as evidenced by their different changes in biomechanics associated with the change in the HAGOS-FSR from pre- to post-rehabilitation.

### 7.4.1 How well can change in biomechanics explain change in HAGOS-FSR?

The ability of the change in biomechanics to explain change in HAGOS-FSR was high across all three clusters, as evident from the large $R^2$ values (0.77-0.93). This is comparable to previous research that used linear models to predict risk factors for ACL injury (Vairo et al., 2008) and ankle lateral ligament sprain (Witchalls et al., 2012). In addition to this, the high cross-validated predicted $R^2$ values obtained (0.74-0.91) provide further confidence in our findings. This demonstrates that on new data, the cluster predictive models generalise well and avoid over-fitting (a common problem in biomechanical research) (Ferber et al., 2016). Furthermore, the ability of the change in biomechanics to explain change in HAGOS-FSR within each cluster was much higher than when examined at a whole group level. In line with previous research (Kienast et al., 1999; Richter et al., 2014), this highlights the importance of examining biomechanical changes at a subgroup (cluster) level.

### 7.4.2 What biomechanical features were used to explain the change in HAGOS-FSR?

With the exception of thorax frontal plane mechanics, there were no clear trend across the three clusters towards a dominance of a single joint or measure for the features retained which may simply highlight the complex neuromuscular changes that occur in response to rehabilitation. However, the strongest predictors in each cluster should perhaps be targeted as part of a rehabilitation programme to improve the HAGOS-FSR (Figure 7.3). In Cluster 1, the strongest predictors for change were thorax contralateral sway velocity, both absolute and relative to the pelvis during the deceleration phase, and ankle plantar flexor concentric work during the concentric phase. In Cluster 2, the strongest predictors were thorax to pelvis contralateral sway velocity in the concentric phase; and knee adductor power absorption and thorax to pelvis ipsilateral sway velocity during the deceleration phase. In Cluster 3, the strongest predictors of a change in HAGOS-FSR were an increase in knee concentric work in the concentric phase, a decrease in thorax ipsilateral sway for the entire ground contact, and an increase in ankle eccentric work done in the early concentric phase.

Interestingly, a strong relationship between a single biomechanical variable and HAGOS when considered in isolation (bivariate correlation) did not necessitate a strong relationship with HAGOS when other biomechanical variable are included in the model (beta coefficient). This is known as Simpson's paradox (Tu, Gunnell and Gilthorpe, 2008), and in the case of ankle plantar flexor eccentric work in Cluster 3, resulted in a reversal of the association between the change in this feature and the change in HAGOS-FSR. This is perhaps not surprising given the inter-segmental nature of human movement and highlights the importance of utilising multivariate models when examining the biomechanics of agility tasks. Surprisingly perhaps, the biomechanical variables on which the clusters were

originally grouped (Franklyn-Miller et al., 2017) were not always included in the statistical models of this study. Furthermore, our study both identified additional potentially important biomechanical changes and did not include all the variables identified by King et al (2018), which only examined pre- to post- changes at a whole group level. These findings cumulatively illustrate the importance of examining biomechanical changes from pre- to post-rehabilitation at a sub-group (cluster specific) level to avoid the potentially masking effects of a whole group analysis (Richter et al., 2014).

### 7.4.3 How can the biomechanical changes be explained?

While we cannot conclude causality between the change in HAGOS-FSR and the underlying biomechanical changes observed in this study, such a relationship could be explained theoretically by at least four possible influencing factors. Firstly, a biomechanical change may represent a reduction in pain. With injury, pain provides an organismic constraint (Hodgesv and Tucker, 2011) and resolution of this pain may result in an alteration of movement technique and/or allow the full utilisation of neuromuscular capacity. The effects of experimental pain on movement is well established (Hodges et al., 2003; Graven-Nielsen and Arendt-Nielsen, 2008; Madeleine and Mathiassen, 2008) and has even been examined with respect to experimental groin pain (Jansen, Poot, et al., 2010). Secondly, biomechanical changes may arise from a change in the velocity of the task, which would influence the kinetics and kinematics of the pivot step. Indeed it is well acknowledged that approach velocity can have a substantial impact on movement mechanics during the running cut (Vanrenterghem et al., 2012; Caekenberghe et al., 2013; Browne and Franz, 2017). Thirdly, a change in task constraints (e.g. an emphasis on keeping the trunk upright during agility tasks) could account for some of the changes observed (Schreurs, Benjaminse and Lemmink, 2017).

For example, it has been demonstrated the trunk mechanics has a large influence on the loading experienced by the lowers limbs both during landing (Blackburn and Padua, 2008) and running cut tasks (Nagano et al., 2011; Frank et al., 2013). Finally, the change in biomechanics may be attributed to neuromuscular training/detraining following rehabilitation. For example Dai et al. (2012) identified that detraining alters knee mechanics during landing increasing the risk of ACL injury while a six week training program was sufficient to enhance performance and reduce risk factors for ACL injury (Myer et al., 2005). Given the high degrees of freedom available in human movement (Bernstein, 1967), any of the above factors could result in a process by which the movement technique is re-optimised during the rehabilitation period.

**Antagonistic biomechanical changes**

One challenge within this present study is in understanding why a number of variables actually dis-improved (an antagonistic change) despite a successful rehabilitation (See Figures 7.4 - 7.6). For example, in Cluster 1 there was a mean increase in ankle concentric work done following rehabilitation, yet both the bivariate correlation and standardised beta score would suggest that such an increase in ankle concentric work is associated with a smaller improvement in the HAGOS-FSR (rather than the desired large change). While, these antagonistic changes could indicate that the biomechanical variables only have an associative (not a causative) relationship with the change in the HAGOS-FSR, this is perhaps unlikely given that all AGP subjects examined in this study returned to pain free participation in sport in a time frame that compares favourably with anatomically-specific rehabilitation protocols (Holmich et al. 1999; A. Weir, Jansen, et al. 2011) and surgical procedures (King et al., 2015). Rather, there are two possible explanations for an antagonistic change within this present study. Firstly, it is possible that a dis-improvement represents a trade off between the contrasting demands of im-

proving task performance and avoiding potential pathomechanics. For example, within Cluster 1 there was a mean increase in thorax contralateral sway velocity in an attempt to position the thorax towards the intended direction of travel. This would be advantageous to change of direction performance because the thorax, head and arms account for almost 68% of body mass (Winter, 2009) and have a strong influence on the orientation of the resultant ground reaction force (Kugler and Janshen, 2010). However, prioritising task performance may overload the commonly painful pubic symphysis region as the internal and external obliques control, at least in part, lateral trunk sway and help form the aponeurosis of the pubic symphysis (Schilders, 2000). This negative association between increased thorax contralateral sway velocity and change in HAGOS-FSR may therefore reflect that those who best improve their HAGOS-FSR require less contralateral trunk adjustment. Conversely, a second explanation for an antagonistic change is that the observed findings are driven by a change in the dynamics of the task itself and not directly related to the pathomechanics of AGP. Within our study, the increase in resultant ground reaction force impulse [which is likely associated with an increase in task execution velocity (Spiteri et al., 2015)] arises at least in part from an increased ability to deliver higher joint kinetics (due to decreased pain or increased neuromuscular capacity) and will likely result in a change in movement technique. This was observed by Browne et al (Browne and Franz, 2017) who independently altered both task velocity and concentric forces during walking and demonstrated that a change in concentric force results in a redistribution of joint contribution. Similarly, during over-ground sprint acceleration (a task similar in demands to change of direction), an increase in acceleration is achieved primarily through increased hip extensor power production (Caekenberghe et al., 2013). Within our study however, rehabilitation coincides with a decrease in trunk flexion (see appendix F), which is negatively associated with contribution of the hip extensors (Blackburn and Padua, 2008; Kopper, Ureczky and Tihanyi, 2012).

Given the contrasting demands between an increase in task execution velocity and less trunk flexion, this may explain the inverse relationship between an increase in ankle concentric work in Cluster 1 and improvements in the HAGOS-FSR, as those unable to utilise there hip extensors effectively, redistribute the demands of the task.

While it is not possible within the scope of this study to separate changes associated with a change in the dynamics of task execution versus those changes related to the pathomechanics of AGP, the high explanatory power of the models along with the poor predicted power of the shuffled models [e.g. applying the model trained on Cluster 1 to the subjects in Cluster 3 (Table 7.5)] allude to potentially important cluster specific biomechanical changes that occur within each AGP cluster from pre- to post-rehabilitation. Future research, should therefore replicate this study whilst controlling for the potentially confounding factors of variable task velocity and/or concentric forces, similar to approaches used when assessing the effect of rehabilitation strategies on linear walking and running (Roper et al., 2016).

**Post rehabilitation classification**

To enhance our understanding of how rehabilitation affects the mechanics of cutting in AGP patients, the AGP subjects were classified post-rehabilitation using a correlation approach outlined in Richter et al. (2017). Interestingly, 48% of subjects within this study actually changed cluster membership when classified post-rehabilitation (Table 7.6). In comparison to clustering which groups subjects together based on common features, classification uses the features of previously identified groups (e.g. the pre-rehabilitation clusters) and assigns each subject to each group (Witten and Frank, 2005). The change in cluster membership may provide an indication of how the AGP patients are responding to rehabilitation. For example those who change cluster membership have altered how they move

post rehabilitation while those who do not change, may utilise the same movement patterns but with smaller magnitudes. Alternatively these changes may simply reflect neuromuscular adaptations in response to rehabilitation. Of particular note however, is that the four key features utilised for classification (hip flexion angle, ankle rotation angle, ankle flexion moment and thorax flexion) did not feature prominently within the statistical models of this current study. This was unexpected, given the cluster specific biomechanical changes identified in our present study. These finding collectively may suggest that the features utilised to classify these AGP patients are associative rather than causative of AGP. However, clearly this requires prospective research to conclusively determine if the features identified have a causative relationship with AGP

## 7.5 Limitations

Given the aim of this paper to quantify the relationship between the change in biomechanics and the change in 'pain and function' following rehabilitation, only those who returned for post- intervention testing were included in this study. Therefore, potentially important information regarding those who do not complete the rehabilitation program and return to sport is not gained from this study. With respect to the statistical models in this study, while the good cross-validated predictive ability of the models in our research provide confidence in this study's findings, it is acknowledged that the relatively small sample sizes within each cluster (n=22-40) may result in inflated beta coefficient confidence intervals. As noted elsewhere, given that velocity of task execution influences joint kinetics (Vanrenterghem et al., 2012; Caekenberghe et al., 2013; Browne and Franz, 2017), not controlling for this may be a potentially confounding factor. Finally, it is worth noting that when registering this study with clinicaltrials.gov, data collection had already commenced with some subjects in this study. The

protocol however, remained unchanged during this period. Furthermore, return to play was omitted from the registration of this study as a secondary outcome measure.

## 7.6 Conclusion

This was the first investigation to examine the relationship between change in biomechanics following rehabilitation and change in an outcome measure in AGP patients, and did so within three clusters groups. While common group level changes were observed in the HAGOS questionnaire in response to rehabilitation, each cluster group demonstrated unique biomechanical features that engendered changes in HAGOS-FSR. In addition, examining the changes that occurred at a sub-cluster level in comparison to a whole group level increased the ability of the predictive models to explain the change in HAGOS-FSR by 81% on average. The predictive models of each cluster are therefore both useful and unique. These findings highlight the importance of examining AGP using a sub cluster approach and reinforce the suggestion that these clusters may benefit from cluster specific targeted rehabilitation. The current study provides a basis for future research to assess the efficacy of such a tailored rehabilitation program.

### 7.6.1 Link between Chapter 7 and Chapter 8

Within Chapter 7, the running cut task was utilised given its ecological validity to actions implicit in the development of AGP. In particular, AGP is common in sports involving repetitive agility tasks (Werner et al., 2009; Orchard, Seward and Orchard, 2013; Thorborg et al., 2017). Given the repetitive loading typical of field based sports, there has been a growing interest in the functional role movement variability may have with respect to overuse injuries such as AGP. In light of this, the aim of Chapter 8 was to investigate if the magnitude of variability differed

between those with and without AGP during multiple repetition of the running cut task.

# Chapter 8

# Is movement variability during a running cut affected by athletic groin pain?

## 8.1 Introduction

Athletic groin pain (AGP) is a common injury, typically associated with sports involving repetitive agility tasks (Werner et al., 2009; Orchard, Seward and Orchard, 2013; Thorborg et al., 2017). Given the association between repetitive loading and chronic overuse injuries such as AGP, there has been a growing interest in the functional role movement variability may have with respect to injury (Nicholas Stergiou and Decker, 2011; Hamill, Palmer and Van Emmerik, 2012; Baida et al., 2017). Movement variability in uninjured individuals represents the natural variation in movement patterns across multiple repetitions of the same task (Bernstein, 1967). However there is divided opinion with respect to the relationship variability has with injury.

Within the scope of dynamic systems theory, it has been suggested too little variability might lead to repetitive loading on a specific tissue structures resulting

in excessive stress and eventual injury (Hamill et al., 1999). In contrast, a recent systematic review identified a trend for injured populations to exhibit greater movement variability when compared to uninjured controls (Baida et al., 2017), while others have theorised that there may be an optimal level of variability (Hamill, Palmer and Van Emmerik, 2012) outside of which there is an increased risk of injury. Within this optimal level of variability theory, a 'U' shaped association characterises the relationship between injury and variability, whereby the risk of injury is higher with either excessive or too little movement variability. Evidence to support this optimal level of variability theory, would suggest the need to quantify normative levels of variability in uninjured populations and target injury rehabilitation on an individualised basis (sufficiently increasing or decreasing variability as appropriate). Furthermore, an optimal level of variability would possibly explain the mixed findings present within the literature with respect to variability and injury (Baida et al., 2017) and highlight the need to use a subgroup analysis when exploring variability. To the best of our knowledge however, no studies have empirically investigated this concept.

In light of the high occurrence rate (Waldén, Hägglund and Ekstrand, 2015), along with the chronic nature and high morbidity of the injury (Thorborg et al., 2017), the examination of movement variability in AGP is warranted. To date however, only one study has specifically investigated movement variability in subjects with and without a history of AGP (Edwards, Brooke and Cook, 2017). Using a running cut task, the authors identified both greater and less variability across the multiple joints and segments examined using magnitude based inference (Batterham and Hopkins, 2006) and concluded that movement variability can distinguish between players with and without a history of athletic groin pain. This is in line with previous research suggesting that often variability can be reduced at one joint whiles increased at another (Miller et al., 2008; Brown, Bowser and Simpson, 2012; Gribbin et al., 2016). It is therefore possible that variability

represents a target for AGP rehabilitation. It is worth noting however, that the small sample size (AGP: n = 7) utilised in this study by Edwards, Brooke and Cook (2016), may render the study's findings unrepresentative of the population. Furthermore, the study utilises discrete points for statistical analysis, which may not adequately capture the variability present across the total waveform (Richter, O'Connor, et al., 2014b; Marshall et al., 2015) and potentially did not utilise an optimal number of cutting trials (see Appendix B). As such the findings from Edwards, Brooke and Cook (2016) need to be substantiated.

The primary aim of this study was to investigate if the magnitude of variability differed between those with and without AGP across the total waveform and secondly if within this cohort there was any evidence of those without AGP exhibiting an optimal level of variability. It was hypothesised that the AGP group would demonstrate significantly greater variability in comparison to the uninjured group and that there would be no evidence supporting an optimal level of variability.

## 8.2 Methods

### 8.2.1 Participants

Twenty AGP patients pre rehabilitation (mean $\pm$ SD: age 23.3 $\pm$ 2.8 yrs., height 1.81 $\pm$ 0.06 m, weight 80.1 $\pm$ 11.1 kg) and twenty recreationally active male field sport athletes (mean $\pm$ SD: age 25.0 $\pm$ 4.9 yrs., height 1.80 $\pm$ 0.05 m, weight 81.23 $\pm$ 6.74 kg) were recruited for this study. All participants were required to between the ages of 18 - 35 and involved, at least recreationally, within multidirectional field sports (see table 8.1). For the AGP group, a primary clinical diagnosis of a pubic aponeurosis injury was made in 11 (55%) cases; hip flexor injury was diagnosed in 3 (15%) cases; adductor injury was diagnosed in 3 (15%) of cases;

Table 8.1: Breakdown of primary sporting participation

| Sport | AGP | | Uninjured | |
|---|---|---|---|---|
| | N= | Percentage | N = | Percentage |
| **Gaelic Football** | 10 | 50 % | 10 | 50 % |
| **Hurling** | 6 | 30 % | 3 | 15 % |
| **Soccer** | 2 | 10 % | 5 | 25 % |
| **Rugby** | 1 | 5 % | 2 | 10 % |
| **Hockey** | 1 | 5 % | 0 | 0 % |

hip joint injury was diagnosed in 2 (10%) cases; and a combined diagnosis of hip flexor injury and hip joint injury was diagnosed in 1 (5%) case. Within this study however, AGP was treated as a single entity (much like lower back pain) and as such inclusion was not restricted by 'entity' (Franklyn-Miller et al., 2017). Additional inclusion criteria required all AGP participants to undergo clinical consultation, MRI imaging and physical examination to confirm diagnosis of AGP as per criteria previously published (Falvey, King and Kinsella, 2015). Exclusion criteria for AGP participants included an underlying medical condition such as inflammatory arthropathy or infection, the presence of hip joint arthrosis, symptoms less than four weeks and a lack of intent to return to pre-injury activity levels. The Sports Surgery Clinic ethics committee approved the study (REF SSC0025) and all of the participants signed informed consent.

### 8.2.2 Biomechanical Model

Twenty-eight reflective markers (14mm diameter) were placed at bony landmarks on the lower limbs, pelvis and trunk as per the Vicon Plug in Gait model (Vicon Motion Systems, Oxford, UK) with an additional marker placed on the anterior aspect of the mid tibia and mid thigh bilaterally. To calculate functional joints, the 'OSSCA' method as implemented in NEXUS 2 was utilised (Taylor et al., 2010). This involves the participant completing a series of movements to estimate the

hip joint centre position and knee rotation axes. To estimate the hip joint centre, the participant completed a star arc pattern with their hip consisting of flexion-extension, abduction-adduction and internal-external rotation while standing on the contralateral limb. To calculate the knee rotation axes the subjects performed a squat movement pattern three times. The hip joint centre and the functional knee axes were then calculated within Vicon Nexus 2 using the symmetrical centre of rotation estimation (SCoRE) (Ehrig et al., 2006) and the symmetrical axis of rotation approach (SARA) (Ehrig et al., 2007), respectively. Soft tissue artefact was minimized using the optimal common shape technique (OCST) (Taylor et al., 2005), where an optimum rigid marker configuration for each segment is formed to reduce the effects of skin elasticity.

### 8.2.3 Data acquisition

Three dimensional marker positions were tracked using 10 infrared cameras (Vicon - Bonita B10, UK), synchronized with two 40x60cm force platforms (AMTI - BP400600, USA) collecting ground reaction force data. Motion and force data were captured at a sampling frequency of 200 Hz and 1000 Hz, respectively. Prior to the experimental testing, the subjects completed a standardized warm-up (Marshall et al., 2015) involving a three-minute jog at a self-selected pace, five body weight squats and two practice trials of the 110° cutting task. For the cut, participants ran as fast as possible toward a marker placed on the floor, made a single complete foot contact on the force plate, and performed a 110° cut with respect to the approach direction before running maximally to the finish (Figure 8.1). In line with findings from Appendix B, seven trials were examined for each subject.

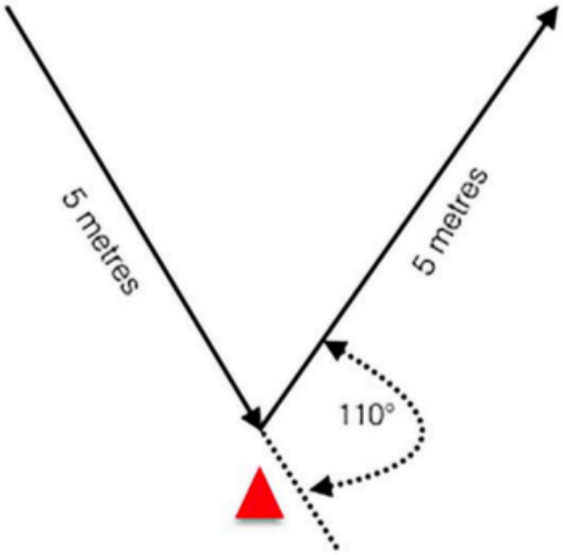


Figure 8.1: Illustration of the 110° cut task

### 8.2.4 Data processing

Both marker and force data were filtered using a fourth order Butterworth filter with a cut-off frequency of 15Hz (Kristianslund, Krosshaug and van den Bogert, 2013). Kinematic and kinetic calculations were performed in Nexus software (Vicon Motion Systems, Oxford, UK) and were subsequently exported to Matlab 2015b (Mathworks, USA) for further processing and statistical analysis. Data from the pivot leg, pelvis and trunk were examined during the pivot step, defined as when the vertical ground reaction force exceeded a 5N threshold. All data was normalised to 101 data points using a cubic spline. Variability was examined for both individual joint/segment angles and also the coordination between every joint/segment angle. A full list of the variability variables examined are presented in table 8.2.

Table 8.2: Summary of Biomechanical variables examined

| Variable | Description of Calculation |
|---|---|
| **Ankle, Knee and Hip joint angle** | Relevant angle between two adjacent segments described with respect to the proximal coordinate system. |
| **Thorax and Pelvis angle** | Absolute angle relative to global coordinate system. |
| **Co-ordination between segments/joints** | Coupling angle calculated from the vector orientation between two adjacent data points on an angle-angle plot relative to the right horizontal. |
| **Resultant impulse** ‡ | $\sqrt{}$ [(Force$_x$ * time)$^2$ + (Force$_y$ * time)$^2$ + (Force$_z$ * time)$^2$ ] |
| **Cutting angle** ‡ | Relative angle between COM displacement vectors calculated incoming (from prior to initial contact to end of the ground contact deceleration phase) and outgoing (from the start of the a acceleration phase to after toe off). |

COM = Centre of mass; ‡ = Manually calculated variable.

### 8.2.5 Co-ordination calculation

To calculate the co-ordination between joints, a modified vector coding approach was utilised (Sparrow et al., 1987). An angle-angle plot between two joint angle time series was constructed (Figure 8.2) and for each data point during the normalised stance phase, the coupling angle ($\gamma$) between consecutive coordinates in the angle-angle plot was calculated to produce values between 0° and 360°:

$$\gamma = [tan^{-1}(\frac{\theta_{y_{i+1}} - \theta_{y_i}}{\theta_{x_{i+1}} - \theta_{x_i}})] \; modulo \; 360 \tag{8.1}$$

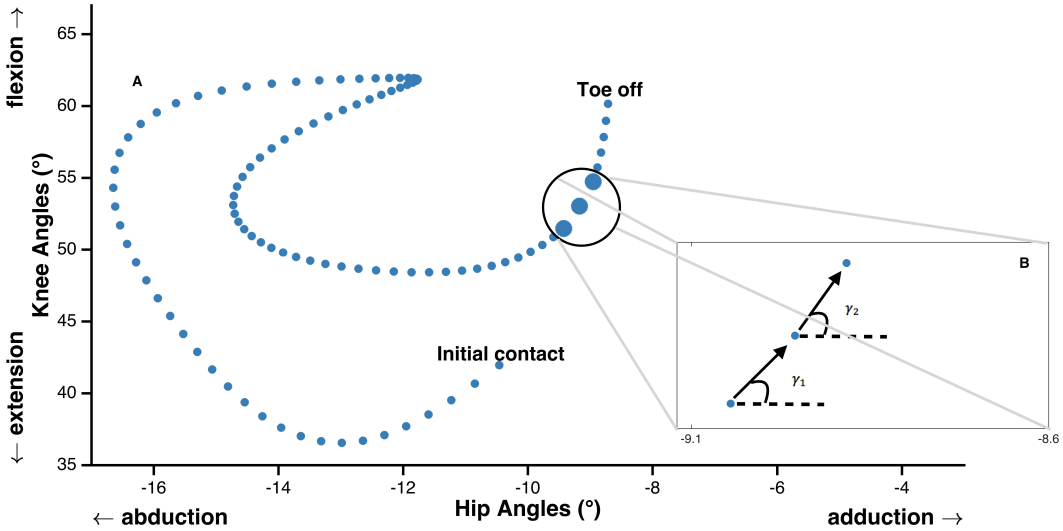


Figure 8.2: A). Angle-Angle plot between knee flexion/extension and hip abduction/adduction. B). Coupling angle $\gamma$ calculated from the vector orientation between two adjacent data points on the angle-angle plot relative to the right horizontal.

### 8.2.6 Variability Calculation

To calculate variability for non- coordination data involving individual joint/segment variables, the between trial standard deviation was calculated at each time point. For the co-ordination data (coupling between any two segments/joints), circular statistics were implemented to account for the directional nature of the coupling angle (Batschelet, 1981). To calculate co-ordination variability, the mean Cartesian co-ordinates from each data point coupling angle plot was calculated:

$$\bar{x_i} = \frac{1}{n}\sum_{i=1}^{n} cos\gamma_i \tag{8.2}$$

$$\bar{y_i} = \frac{1}{n}\sum_{i=1}^{n} sin\gamma_i \tag{8.3}$$

These values were then utilised to calculate the mean vector length.

$$\bar{r_i} = \sqrt{(\bar{x}_{i^2} + \bar{y}_{i^2})} \tag{8.4}$$

This then allows for the circular equivalent of standard deviation ($AD_i$) to be calculated.

$$AD_i = \sqrt{2 * (1 - \bar{r}_i)} * \frac{180}{\pi} \tag{8.5}$$

### 8.2.7 Optimal variability investigation

A secondary aim of this study was explore the possibility of an optimal level of variability. Given the non-negative nature of variability, all data in both the uninjured group and the athletic groin pain group were firstly normalised by subtracting the mean of the uninjured group. This would theoretically centre both groups on zero normalised variability, with (if the optimal variability theory proved true) the AGP group further from zero than the uninjured group (Figure 8.3).

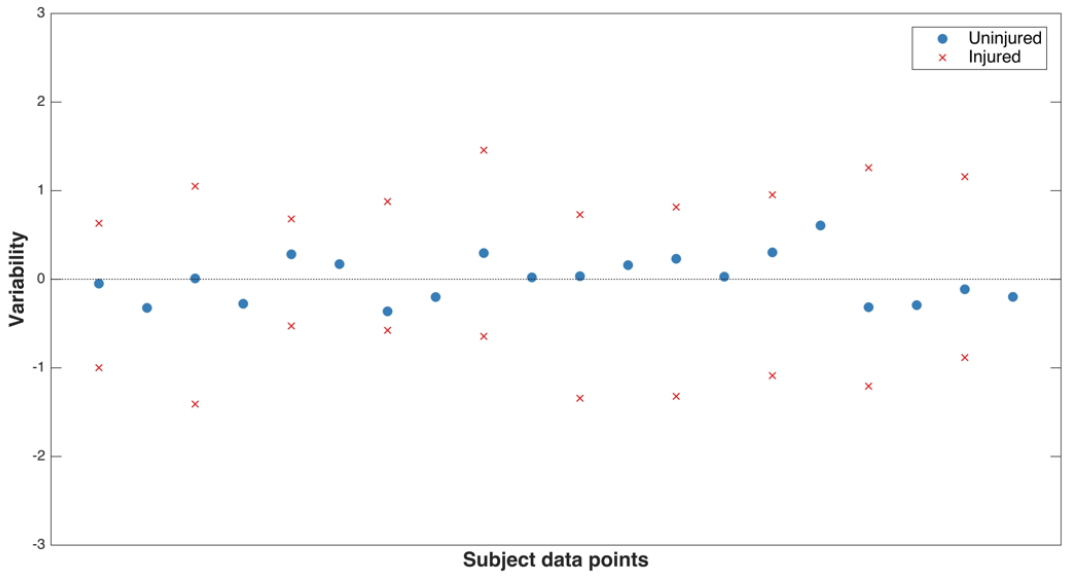


Figure 8.3: Proposed 'Optimal variability' data distribution normalised so all data would centre on the zero mean of the uninjured group.

To allow for statistical comparison of this pattern, the absolute normalised variability was calculated, thus making the data positive whilst preserving the ab-

solute distance from zero normalised variability (Figure 8.4). This was completed for each time point $i$.

$$AD_{norm_i} = \sqrt{[AD_i^2 - \bar{AD}_i^2(uninjured)]} \tag{8.6}$$

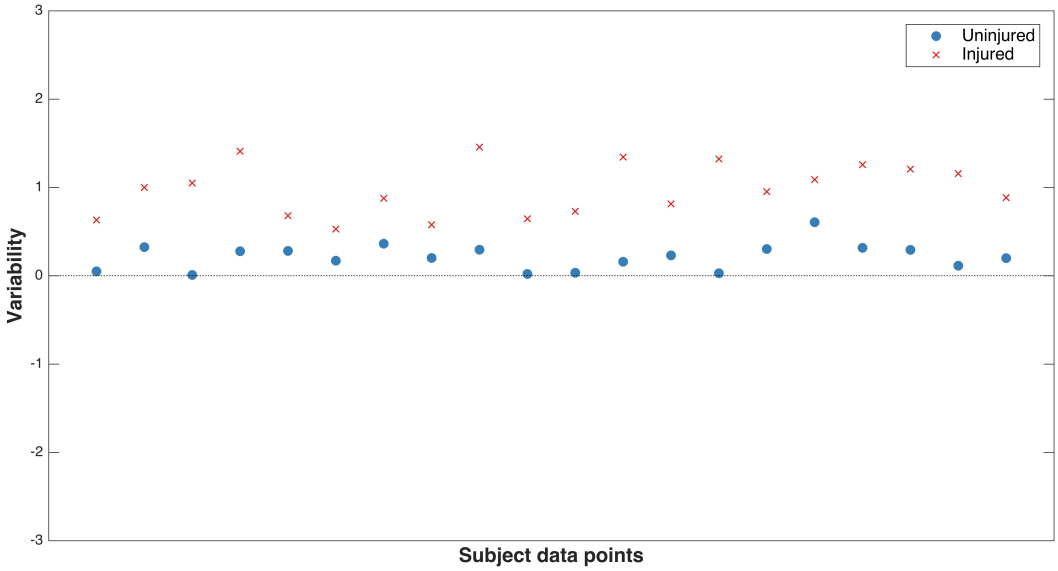


Figure 8.4: Absolute normalised variability data

### 8.2.8 Statistical Analysis

Given the positively skewed nature of variability data at a population level (Privitera, 2015), to be robust to assumptions of Gaussian distributions, non-parametric statistics were utilised to examine differences between the AGP group and the uninjured group. To statistically test the whole waveform, a curve analysis was performed using one-dimensional statistical non-parametric mapping which utilises a non-parametric permutation test (Nichols and Holmes, 2002). Given the computational expense of testing all possible permutations, 10,000 random permutations were chosen. To control for within feature/variable family wise error, the significance of the statistical non-parametric mapping curves were determined topologically using random field theory (Adler and Taylor, 2007). To

ensure our findings were robust to noise in the data and the influence of a single subject, two steps were taken. Firstly only phases of duration 3% or longer were considered as true differences (Gribbin et al., 2016). And secondarily, a 'leave one out' approach was taken in which every combination of n and n - 1 subjects from the AGP group were compared to every combination of n and n - 1 subjects from the uninjured group (441 different comparisons). A heat map was then generated and only phases that were significantly different $>$ 95% of the time were retained (Figure 8.5). All statistical analyses were conducted in Matlab 2015b (Mathworks, USA) and the open source package: spm1d (ver. 0.4 - www.spm1d.org). Cutting angle and eccentric impulse were statistically tested using independent t tests. The alpha level was set at 0.05 and no corrections for multi between variable comparisons were made (Perneger, 1998; Hopkins et al., 2009). All results are presented as mean $\pm$ SD. Rank biserial correlation effect size was reported as weak (0.2 - 0.5), moderate (0.5 - 0.8) and strong ($>$ 0.8) (Ferguson, 2009).

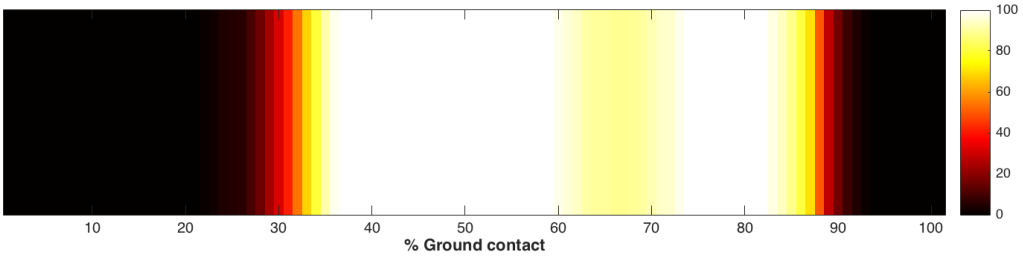


Figure 8.5: Example significance heatmap for pelvis flexion/extension angle. Colour bar indicates the percentage of simulations where significane was identified. White phases are retained as 'stable phases.'

## 8.3 Results

There was no significant difference in cutting angle (AGP: 106.5 $\pm$ 5.4°, Uninjured: 104.4 $\pm$ 0.09°, p =0.23) or resultant eccentric impulse (AGP: 0.51 $\pm$ 0.08 N.s.kg$^{-1}$, Uninjured: 0.55 $\pm$ 5.3 N.s.kg$^{-1}$, p =0.19) between the two groups.

### Individual joint/segment variability

For individual joint/segment angles, pelvis flex/extension variability was significantly greater in the AGP group in comparison to the uninjured group over two phases, while for ankle plantar/dorsi flexion variability the AGP group had significantly less variability in comparison to the uninjured group (Table 8.3).

### Inter joint/segment Co-ordination variability

For co-ordination variability, 'thorax ab/adduction - hip flexion/extension' coupling, and 'pelvis ab/adduction - hip flexion/extension' coupling variability were significantly different with the AGP demonstrating significantly less variability in comparison to the uninjured group (Table 8.3).

### Optimal variability

When exploring the possibility of an optimal range of variability, this study identified no significant differences between the AGP and uninjured group for any of the measures investigated.

Table 8.3: Significant variability findings

| Variable | Percent | AGP | Uninjured | Sig | RBC | Diff |
|---|---|---|---|---|---|---|
| **Thorax abd/ad - Hip flex/ext** | 88-93% | 10.99 ± 8.54 | 27.21 ± 13.28 | <0.01 | 0.65 | < |
| **Pelvis abd/ad - Hip flex/ext** | 89-91% | 10.85 ± 7.50 | 24.83 ± 1342 | <0.01 | 0.57 | < |
| **Ankle dor/plan flexion** | 79-93% | 3.74 ± 1.28 | 5.8 ± 2.01 | <0.01 | 0.51 | < |
| **Pelvis flex/ext** | 36-63% | 4.42 ± 1.66 | 2.86 ± 1.38 | <0.01 | 0.49 | > |
| **Pelvis flex/ext** | 70-84% | 4.02 ± 1.52 | 2.66 ± 1.21 | <0.01 | 0.49 | > |

Dor/Plan = Dorsi/plantar, Ab/add = abduction/abduction, flex/ext = flexion/extension, Sig = significance (p), RBC = rank-biserial correlation, Diff = AGP direction of difference relative to the uninjured group.

## 8.4 Discussion

Rather than noise or error, it has been suggested that variability represents the natural variations in movement from one task to the next and may have an important role to play with respect to reducing repetitive loading and ultimately risk of injury (Hamill et al., 1999). This study sought to investigate if movement variability during a planned 110° cutting task was affected by athletic groin pain. This study identified four variables whose variability differed significantly between the AGP group in comparison to the uninjured group. Three variables demonstrated reduced variability (Thorax ab/adduction - hip flexion/extension coupling, pelvis ab/adduction - hip flexion/extension coupling, and ankle dorsi/plantar flexion angle) with moderate effect size (RBC = 0.51 - 0.65), while only pelvis flexion/extension angle variability was significantly greater in the AGP group (RBC = 0.49).

The trend identified within this current study that reduced variability is associated with injury is in opposition to the conclusions from a recent systematic review which suggested that injured subjects have a tendency to have greater variability when compared to uninjured controls (Baida et al., 2017). It is unclear why this current study found opposing evidence to the systematic review, since as per the recommendations of the review we examined a non-cyclic task utilising a continuous waveform approach (Baida et al., 2017). Differences however may be related to AGP affecting variability differently than other forms of injury examined in the review or perhaps the studies within the review not examining an optimised number of trials (see appendix B).

In contrast however, the findings from this present study are supported by the trend identified by the only other study that has specifically examined variability in those with and without a history of AGP (Edwards, Brooke and Cook, 2017). Using an unanticipated cutting task, the authors found that 82% (48/59) of the

variables examined were substantially different between the two groups with 65% (31/48) of these variables demonstrating less variability in those with a history of AGP in comparison to uninjured controls.

In comparison to the study of Edwards, Brooke and Cook, (2016), clearly this present study identified far fewer variables which differed between those with and without AGP. There are two main factors that may in part explain this. Firstly, Edwards, Brooke and Cook, (2016) utilised a magnitude based inference statistical approach which is a lot less conservative than null hypothesis testing with an alpha level of 0.05 (as utilised within in this current study). This potentially may have lead to a high false positive rate in Edwards, Brooke and Cook, (2016). Secondly and perhaps more importantly, the study by Edwards, Brooke and Cook, (2016) examined only 7 AGP subjects, which may render their findings unrepresentative of the population and hazards the possibility that their findings were influenced by outliers. In contrast, this current study examined 20 subjects per group and utilised a 'leave one out' technique to ensure the results were not affected by a small number of subjects. Interestingly, when reviewing the study by Edwards, Brooke and Cook, (2016) for commonalities to our study, the only common feature identified was less variability in ankle dorsi/plantar flexion in AGP patients in comparison to uninjured controls.

There are at least four possible interpretations for the results of this study. Firstly, where lower movement variability was observed in the AGP group in comparison to the uninjured controls, it is possible that this may represent a risk factor for AGP (rather than an outcome of the injury). In line with the dynamic systems theory of injury (Hamill et al., 1999) and current thoughts that AGP arises from a chronic overload of the pubic region (Meyers, Greenleaf and Saad, 2005; Verrall et al., 2007; Franklyn-Miller et al., 2017; King et al., 2018), the lower variability may have resulted in repetitive loading to the pubic symphysis region and ultimately injury. Secondly, as a compensatory mechanism, the pain or injury

associated with AGP may be causing an organismic constraint (Newell and Slifkin, 1998). In order to avoid loading on the painful structures, the AGP may reduce variability by restricting technique to limit potentially painful motion. The notion of a compensatory mechanism was also previously suggested by Heiderscheit, Hamill and Emmerik, (2002), who identified reduced variability in patients with patella femoral pain syndrome that subsequently re-normalised under a taping treatment. Thirdly, it is possible that variability is reduced in the AGP group in comparison to the uninjured group in response to the detraining that commonly occurs with the injury (Hölmich and Thorborg, 2014). It is theorised that variability represents the exploration of new movement strategies and solutions (van Emmerik and van Wegen, 2002). However, in line with the uncontrolled manifold hypothesis, it appears that the human body may only utilise variability as an exploratory means once achievement of the outcome goal (the change of direction task) is not threatened (Scholz and Schöner, 1999). Thus, drawing parallels to literature comparing young and old subjects (van Emmerik and van Wegen, 2002), detraining may have increased the relative demands of the cutting task for the AGP group, and as a result, the AGP group may prioritise task achievement over the exploration of new movement strategies. Finally, within this study, for one variable (pelvis flexion/extension), variability was actually greater in the AGP group in comparison to the uninjured group. Clearly this observation can similarly be explained as a risk factor, a compensatory mechanism or as a response to detraining (Baida et al., 2017), however, it is unclear why a group can simultaneously have both greater and less variability across joints and segments. It is worth note that this pattern of both greater and less variability across joints is not uncommon and has been observed in studies investigating chronic ankle instability (Brown, Bowser and Simpson, 2012), iliotibial band syndrome (Miller et al., 2008) and AGP (Edwards, Brooke and Cook, 2017).

Interestingly the four segments/joints contributing to the significant findings (thorax, pelvis, hip and ankle) have all been theorized to play a role in the pathomechanics of AGP (Cowan et al., 2004; Sayed Mohammad, Ragaa Abdelraouf and Abdel-Aziem, 2013; Franklyn-Miller et al., 2017; King et al., 2018). Given the close proximity of the pelvis and hip to the region of AGP injury it is was not unexpected that these features would feature prominently in the results of this current study, since AGP is commonly considered to be caused by a repetitive overload to the pubic symphysis region and surrounding structures (Meyers, Greenleaf and Saad, 2005). Indeed hip extensor mechanics were identified as a potential target for rehabilitation in both Chapter 3 and 5 investigating the biomechanics of a lateral hurdle hop task.

Similarly, the thorax, is controlled at least in part by abdominal muscles that join the tendinous fascia of the common adductors inferiorly to form an aponeurosis of the pubic symphysis (Robertson et al., 2009). Accounting for almost 68% of body mass (Winter, 2009), the position of the thorax can also have a large influence on the loading experienced by the lower limbs and pelvis (Blackburn and Padua, 2008; Kopper, Ureczky and Tihanyi, 2012; Frank et al., 2013; Sasaki et al., 2015). The importance of thorax frontal mechanics to the pathomechanics of AGP has previously been highlighted in Chapter 7 and in previous research by King et al., (2018).

Finally, while the importance of the ankle joint to AGP is less intuitive than other proximal segments, it is important to note that the ankle is very important for the modulation of stiffness and load absorption during dynamic tasks (Farley and Morgenroth, 1999; Lewis and Ferris, 2008; Hobara, Muraoka and Omuro, 2009; Yeow, Lee and Goh, 2011). Furthermore the ankle has almost consistently been identified as a variable of interest with respect to AGP across studies (Edwards, Brooke and Cook, 2017; Franklyn-Miller et al., 2017; King et al., 2018) and within this thesis (Chapter 3, 4, 5 and 7). As such, it is suggested that in accordance

with previous research (Edwards, Brooke and Cook, 2017) increasing variability may represent a target for AGP rehabilitation. While no research has specifically examined how rehabilitation exercises affect movement variability at a joint level, it is suggested that as per dynamic systems theory (Newell 1986), rehabilitation aiming to increase movement variability should not strive for an ideal movement, but should facilitate the emergence of whole body movement solutions through the interaction of patient, task and environment constraints (Harbourne and Stergiou, 2009; Lee et al., 2014). For example this may involve landing on an unstable surface to force the exploration of new movement solutions to landing.

A secondary aim of this study was to determine if within this cohort there was any evidence of those without AGP exhibiting an optimal level of variability. Previously it has been suggested that in line with dynamic systems approach to injuries and movement variability, that there is an optimum range of variability that exists for human movement, outside of which there is an increased risk of injury (Stergiou, Harbourne and Cavanaugh, 2006; Hamill, Palmer and Van Emmerik, 2012). In line with this theory, there should exist a 'U' shaped association in the relationship between injury and variability, whereby injured populations would lie at the extremes of this association (excessive or too little movement variability), while uninjured populations would lie in the middle of this continuum. This would possibly explain the mixed findings present within the literature with respect to variability and injury (Baida et al., 2017). To the best of the authors' knowledge however, this is the first study to empirically test this hypothesis and was unable to detect any indication of an optimal variability pattern within this population and task examined. This suggests that rehabilitation aiming to affect movement variability does not need to be tailored to the individual's variability findings. However, clearly further research utilising large sample sizes is required to refute or confirm the findings of this current study.

### 8.4.1 Limitations

There are two main limitations associated with this research. Firstly, a major limitation of the current study is the number of between variable comparisons made (66). While we attempted to increase the robustness of our findings by retaining phases only 3% or longer (Gribbin et al., 2016), and utilising a 'leave one out' technique, we acknowledge that there may be a high false positive rate. While research often controls for multiple between variable comparisons (using for example a Bonferroni correction), doing so within this exploratory research would result in a p-value unrealistically stringent and subsequently increase the risk of type two errors (Perneger, 1998; Hopkins et al., 2009). Rather we believe that in light of the findings from this study and the trends exhibited by Edwards, Brooke and Cook, (2016), future research is warranted to confirm the findings of this study. A second possible limitation was that this current study utilised seven trials following findings from previous research (see Appendix B). Within Appendix B the number of trials required for this current study was determined by identifying the trial number that maximised the observed effect sizes. While the authors believe that this is methodologically correct, it is acknowledged that utilising the same cohort for both the number of trials study (Appendix B) and the present comparison study may have been biased by increasing our likelihood of identifying differences.

Future research should replicate this present study to confirm if the four identified features in this study are truly affected by AGP rather than artifacts of multiple comparisons. Furthermore, the replicating study should utilise the number of trials identified in Appendix B, but on a new sample.

## 8.5 Conclusion

The results from this study indicate that AGP patients may be characterised by reduced movement variability in comparison to uninjured controls for three out of the four significantly different variables identified. Should this be the case then variability may represent a target for AGP rehabilitation. Given the large number of comparisons made however, no clear-cut conclusions can be drawn from this research. Future confirmatory research, using appropriate methodology is warranted to ascertain if variability of the features identified in this current study are truly affected by AGP or are a product of multiple comparisons. Further, prospective research is ultimately required to conclusively determine if altered movement variability increases the risk of injury.

# Chapter 9

# General Discussion

The primary aim of this PhD thesis was to evaluate the biomechanical factors affected by AGP to enhance our understanding of this injury. This was conducted across five experimental studies and utilising two test actions, namely the lateral hurdle hop and the running cut. This concluding chapter will synthesize the findings from this PhD thesis and broadly categorise the experimental studies for discussion into research investigating the biomechanical magnitude domain (Chapters 3, 4, and 7 ) and research into movement complexity and variability (Chapters 5 and 8).

## 9.1 Biomechanical magnitude domain

The results from this thesis have identified multiple biomechanical factors that are affected by AGP. In particular it would appear that the thorax, hip and ankle were commonly identified across the experimental studies. Given the proximity of the hip joint to the region of pain, it is not surprising that this joint was affected by AGP. Within Chapter 3 & 4 hip abductor stiffness and moments were less in the AGP group pre- rehabilitation in comparison to the uninjured group and changed significantly with rehabilitation, suggesting that increasing hip abductor moments

and hip abductor stiffness are potential targets for rehabilitation. These findings are supported by previous research by both King et al., (2018), who identified a reduction in hip adductor moments from pre- to post- AGP rehabilitation during a running cut task, and research by Morrissey et al., (2012) who demonstrated reduced gluteus medius activity in AGP patients in comparison to uninjured controls.

With an upright trunk during single leg stance, the ground reaction force passes medially to the hip joint centre (Fig 9.1). This posture is typical during the lateral hurdle hop test (Marshall et al., 2015) and requires the hip abductors to act eccentrically to control the pelvis while producing a concomitant increase in ipsilateral hip joint reaction forces (Neumann, 2010; Wesseling et al., 2015).

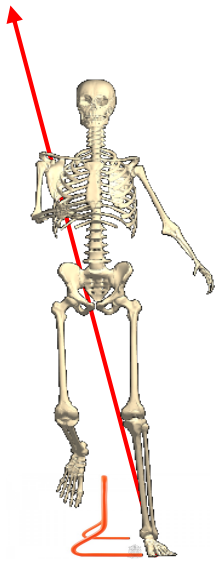

Figure 9.1: Illustration depicting the hurdle hop task

Reduced hip abductor moments/stiffness in those with AGP pre- rehabilitation (Chapter 3 & 4) may therefore be utilized as a compensatory mechanism to reduce hip joint loading since hip trauma can result in a referral of pain to the medial thigh region (Lesher et al., 2008). In a similar light, ipsilateral trunk sway may be used to reduce the external hip adductor moment arm, thereby offloading the hip abductors and hip joint (Neumann, 1989).

In contrast, within the literature the most consistently identified clinical factor associated with AGP was hip adductor weakness (Malliaras et al., 2009; Crow et al., 2010; Nevin and Delahunt, 2014). As a result of this, exercise interventions have generally targeted hip adductor strength (Holmich et al., 1999; Weir et al., 2011). It is possible however, that hip adductor weakness is an associative change due to neural inhibition in response to pain rather than a risk factor for AGP. Indeed, while Crow et al., (2010) identified prospectively hip adductor weakness prior to the onset of AGP, this was identified no earlier than two weeks prior to injury, suggesting that the change in adductor strength was associated with subsensory pain rather than a true risk factor. Similarly, despite the absence of any isolated hip adductor strengthening in both this thesis and research by King et al., (2018), hip adductor strength increased significantly from pre- to post-rehabilitation suggesting that this may be related to the reduction of pain.

During tasks where trunk posture is less controlled than in the hurdle hop (such as during the running cut), the trunk also featured prominently in the studies' findings. Indeed, in Chapter 7 where the running cut was examined, the only consistent and strong predictor within each of the clusters was thorax contralateral sway velocity, suggesting that thorax contralateral sway position and velocity should be targeted with rehabilitation. The importance of the trunk to AGP mechanics is supported by previous research by Edwards, Brooke and Cook, (2017) and King et al., (2018) who both also found that the thorax was affected by AGP during running cut tasks. The prevalence of studies finding that trunk mechanics is affected by AGP is not surprising given that the thorax (including arms and head) accounts for almost 68% of body mass (Winter, 2009), and can have a large influence on the loading experienced by the lower limbs and pelvis (Blackburn and Padua, 2008; Kopper, Ureczky and Tihanyi, 2012; Frank et al., 2013; Sasaki et al., 2015).

In contrast to the hurdle hop task where a moderate ipsilateral lean can act to offload the hip abductors and hip joint, during a running cut task excessive ipsilateral trunk sway may increase the hip adductor moment which acts to eccentrically control the trunk and pelvis (Hewett and Myer, 2011) (Fig 9.2).

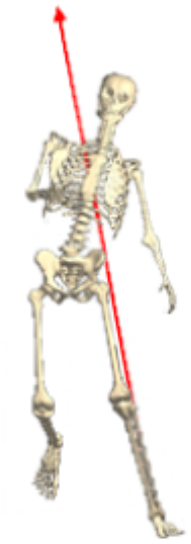

Figure 9.2: Illustration depicting the running cut task

This may ultimately result in an increase in sheer stress at the pubic symphysis, resulting in pain and possibly neural inhibition of this commonly weakened muscle. Additionally, the anatomical continuity between the proximal adductor longus tendon, the pubic symphysis anterior capsule and the distal rectus abdominis attachment, provide a mechanism for which suprapubic pain can be experienced with increased tensile force from the adductors (Norton-Old et al., 2013). The association between hip adductor moments and trunk lean was observed by King et al (2018) who found a decrease in hip adductor moments associated with an increase in trunk contralateral lean from pre- to post- AGP rehabilitation.

In addition to affecting the frontal plane of the hip, AGP also appeared to affect the sagittal and transverse planes of this joint. Within Chapter 3, from

pre- to post- rehabilitation there was an increase in hip external rotation power production and hip extensor moments while there was reduction in knee extensor moments. This increase in hip extensor moments from pre- to post- rehabilitation was observed despite a significant decrease in thorax flexion angle from pre- to post-rehabilitation (Figure 9.3) suggesting that the hip extensor muscles are working more post-rehabilitation (Kopper, Ureczky and Tihanyi, 2012).

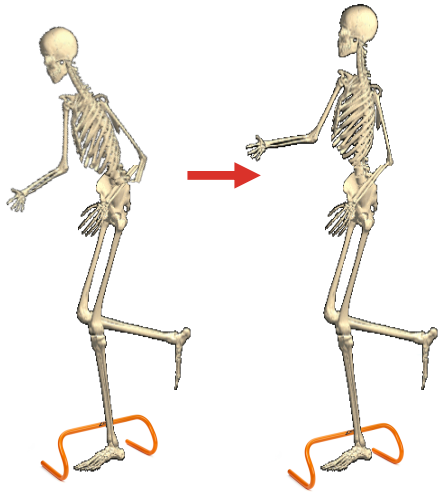

Figure 9.3: Illustration depicting the change in thorax flexion angle from pre- to post- rehabilitation.

This is in contrast to research by King et al., (2018) who identified a decrease in hip extensor moments from pre- to post- rehabilitation during a running cut task. These contrasting findings may indicate that rather than a risk factor for AGP, the change in hip extensor moments from pre- to post- rehabilitation is more related to the execution of the hurdle hop task and the need to have an upright trunk at the time of takeoff. Despite the increase in hip joint moments within Chapter 3, hip extensor power dissipation remained significantly less in the AGP group post-rehabilitation in comparison to uninjured controls. This alludes to a reduced joint angular velocity and potentially a protective mechanism utilized to reduce loading at the hip/pelvic region. This is supported by research by Severin et al.,

(2017) who found decreased angular velocity at the pelvis and hip (stance and swing leg) in those with AGP in comparison to controls during a kicking task.

In addition to AGP affecting the hip and thorax, the ankle joint also featured frequently within the studies of the thesis (Chapter 3,4 & 6). While the role of distal joints in the pathomechanics of proximal injuries is often overlooked, the ankle has a fundamental role to play with respect to modulating whole body stiffness and load absorption during dynamic tasks (Farley and Morgenroth, 1999; Lewis and Ferris, 2008; Hobara, Muraoka and Omuro, 2009; Yeow, Lee and Goh, 2011). It is plausible therefore that if AGP is a result of poor lateral hip/trunk control, the lower moments and stiffness observed at the ankle (and generally in the sagittal plane of the knee and hip) may represent a compensatory mechanism to reduce the magnitude of the resultant ground reaction force and subsequently reduce the external hip abductor/adductor moment. Clearly however, this requires further investigation.

### 9.1.1 Sub-group analysis

A secondary aim of this thesis was to quantify the relationship between the change in biomechanics and change in 'pain and function' in three AGP cluster groups following a common rehabilitation. Previously, our research group had demonstrated for the first time that AGP patients exhibit three distinct biomechanical sub-clusters during a running cut task and suggested that rehabilitation should be specifically tailored for each of these clusters (Franklyn-Miller et al., 2017). However, the authors provided no evidence to suggest that the clusters would respond differently to the same rehabilitation program. The results from Chapter 7 illustrate that each cluster demonstrated distinct changes in response to rehabilitation and that the change in biomechanics within all three clusters had a high ability to explain the associated change in 'pain and function' within that cluster.

Furthermore, the ability of the change in biomechanics to explain change in 'pain and function' within each cluster was much higher than when examined at the whole group level. In line with previous research (Kienast et al., 1999; Richter et al., 2014), this highlights the importance of examining biomechanical changes at a subgroup (cluster) level and provides evidence to support the suggestion that AGP rehabilitation should target each sub cluster.

## 9.2 Variability and complexity

Athletic groin pain (AGP) is prevalent in sports involving repetitive agility tasks (Werner et al., 2009; Orchard, Seward and Orchard, 2013; Thorborg et al., 2017). Given the association between repetitive loading and chronic overuse injuries such as AGP, there has been a growing interest in the functional role movement variability may have with respect to injury (Stergiou and Decker, 2011; Hamill, Palmer and Van Emmerik, 2012; Baida et al., 2017). Movement variability can be assessed in terms of both the amount of variability and the structure of variability (complexity).

Within this thesis, the amount of variability (herein simply referred to as 'variability') was examined in the form of a systematic review (Baida et al., 2017) and as an experimental examination of AGP in comparison to uninjured controls during a running cut task (chapter 8). The results from the experimental study demonstrated a trend for variability to be reduced with AGP for three of the four significantly different biomechanical variables examined (thorax ab/adduction - hip flexion/extension coupling, pelvis ab/adduction - hip flexion/extension coupling, and ankle dorsi/plantar flexion angle). This general trend of reduced variability was in accordance with the trend identified in the only other study which has examined variability in AGP (Edwards, Brooke and Cook, 2017). Interestingly however, the findings were opposition to the conclusions from the

systematic review which suggested that injured subjects have a tendency to have greater variability when compared to uninjured controls (Baida et al., 2017). It is unclear why chapter 8 found opposing evidence from the systematic review, but may be related to AGP affecting variability differently than the other forms of injury within the review or perhaps the studies within the review not examining an appropriate number of trials (see Appendix B).

In a similar light to the lower variability identified in AGP patients in comparison to uninjured controls (chapter 8), complexity was examined within chapter 5 and was identified to be generally reduced in those with AGP. This is in accordance with the general consensus in the literature which suggests that complexity is reduced with injury and pain (Georgoulis et al., 2006; Sndergaard et al., 2010; Tochigi et al., 2012; Terada et al., 2015).

The results from this thesis suggest that both variability and complexity can distinguish between AGP subjects and uninjured controls. In fact, in comparison to more traditional measures, when comparing the AGP group pre-rehabilitation to uninjured controls during a hurdle hop test, complexity demonstrated larger effect sizes (Chapter 5, D = 0.31 - 0.96) than traditional kinematic and kinetic measures (Chapter 3, D = 0.29 - 0.75) or measures of stiffness (Chapter 4, D = 0.36 - 0.79). Further, within this thesis, the combination of the running cut with the examination of variability also achieved moderate effect sizes between those with and without AGP [Chapter 8, (rank biserial correlation = 0.49 - 0.65)]. It is worth noting however, that it is not possible to directly compare complexity and variability in terms of their ability to delineate between those with and without AGP, since within this thesis, both measures were not utilised on the same experimental task. Collectively, these results suggest that both variability and complexity may be utilised to screen for predisposition of AGP and possibly as a more effective outcome measure following rehabilitation than currently

utilised methods. This latter point is particularly pertinent in AGP given the relatively high rate of re-injury (Hagglund, 2006).

Within the chapter exploring complexity in AGP (chapter 5 ), moments had the largest ability to delineate between those with and without AGP. Further, complexity of the hip joint and the sagittal plane of the lower limb joints had the greatest between group differences. Within chapter 8, the sagittal plane was involved in every significantly different variable identified. Interestingly, within both chapters, the sagittal plane of both the hip and the ankle had the largest between group effect sizes.

There are three possible interpretations for the reduced variability and complexity observed in AGP group in comparison to the uninjured controls. Firstly, the reduced complexity and variability in the AGP group may represent a risk factor for AGP. In line with the dynamic systems theory of injury (Hamill et al., 1999) and current thoughts that AGP arises from a chronic overload of the pubic region (Meyers, Greenleaf and Saad, 2005; Verrall et al., 2007; Franklyn-Miller et al., 2017; King et al., 2018), the lower variability and complexity may have resulted in an inability to respond to perturbations that commonly occur during field-based sports, resulting in repetitive loading to the pubic symphysis region. Secondly, it is possible that the AGP group are utilising a more regular, rigid, motor behaviour in an attempt to avoid pain (or perceived threat of pain) which is acting as an organismic constraint (Newell and Slifkin, 1998). This association between pain and reduced complexity has been demonstrated with respect to seated postural discomfort (Søndergaard et al., 2010). Finally, it is possible that complexity and variability is reduced in the AGP group in comparison to the uninjured group in response to the detraining that commonly occurs with the injury (Hölmich and Thorborg, 2014). It is theorised that variability represents the exploration of new movement strategies and solutions (van Emmerik and van Wegen, 2002). However, in line with the uncontrolled manifold hypothesis (Scholz and Schöner,

1999), detraining may have increased the relative demands of the cutting task for the AGP group, and as a result, the AGP group may prioritise task achievement over the exploration of new movement strategies. The effects of detraining on complexity were previously demonstrated in the physiology literature, where heart rate complexity was reduced after just four weeks of detraining (Heffernan et al., 2007).

Considering the findings within this thesis, it is possible that rehabilitation should aim to increase movement complexity and variability in AGP patients to uninjured magnitudes. While traditional methods of rehabilitation (e.g. resistance and balance training) appear to appropriately increase complexity of various physiological systems (Heffernan et al., 2007; Wayne and Kaptchuk, 2008; Millar and Levy, 2013), interventions aiming to restore movement complexity and variability may benefit from following a dynamic systems theory approach (Newell 1986). Such an approach would prioritise variable practice, target multiple systems and facilitate the emergence of whole body movement solutions through the interaction of patient, task and environment constraints (Harbourne and Stergiou, 2009; Lee et al., 2014). Rehabilitation may also benefit from 'over-loading' variability by emphasising the use of a large number of degrees of freedom. This may encourage the exploration of new movement strategies by the neuromuscular system to re-optimize movement in light of altered neuromuscular capacity (e.g. reduced muscular strength) and/or in the presence of pain induced changes to the body schema (Schwoebel, 2001). While this rehabilitation should be conducted utilising whole body movements, research should investigate which rehabilitation exercises best target the variability/complexity at the joints affected by AGP, particularly the sagittal plane of the hip and ankle.

## 9.3 Summary and Future Directions

The aetiology of AGP is likely to be multifactorial given the complexity of this anatomical region (Robertson et al., 2009) and distinct sub-clusters identified within Chapter 7. Despite this, across the five experimental studies of this thesis, the thorax and hip segments appear to be most commonly affected by AGP, particularly in the frontal plane. It is suggested that rather than focusing rehabilitation on the strength of the adductors alone, increasing hip abductor moments/stiffness and enhancing thorax control may reflect targets for AGP rehabilitation. The ankle joint was also consistently and strongly identified as being affected by AGP but generally failed to change significantly from pre- to post-rehabilitation within Chapter 3 & 4. Given the ability of the ankle joint to both absorb load and modulate whole body stiffness (Farley and Morgenroth, 1999; Lewis and Ferris, 2008; Hobara, Muraoka and Omuro, 2009; Yeow, Lee and Goh, 2011), it is plausible that the lower ankle sagittal plane moments/stiffness represents a compensatory mechanism to reduce the magnitude of resultant ground reaction force and offload the hip abductors/adductors. Future research is warranted however, to ascertain if AGP rehabilitation could be further enhanced with an additional focus on improving ankle function. The findings from Chapter 7 reinforce the suggestion that clusters may benefit from cluster specific rehabilitation and the factors identified within each cluster provide a basis for future research to assess the efficacy of such a tailored exercise program.

The results from this thesis also suggest that movement complexity and variability can delineate between those with and without AGP. In particular, it appears that complexity achieved greater effect sizes when comparing AGP patients to uninjured controls than other commonly utilised biomechanical measures. Future research should explore the utility of these metrics to screen for the predisposition of AGP and/or as an outcome measure following rehabilitation. In addition,

research should explore the efficacy of targeting variability and complexity in AGP rehabilitation. This rehabilitation would follow a dynamic systems approach with whole body movements, variable practices and utilising large degrees of freedom (Newell 1986).

Finally, while this PhD thesis has identified biomechanical variables that may be affected by AGP, ultimately prospective research is required to support or refute the findings of this thesis.

# Appendix A

# Appendix A: Additional Information from the variability systematic review

Table A.1: All variables measured and significant findings

| Author | Variables Examined | Significant Finding |
|---|---|---|
| Van Uden et al. 2003<br>ACLR | knee and ankle sagittal plane angular displacements | > operated limb, sagittal plane, ankle-knee coupling (p=0.001) |
| Cordeiro et al. 2015 ACLR | Knee sagittal plane kinematics: ROM, angular velocity, angular acceleration, angular position at maximum velocity. Temporal: duration time to peak velocity and to peak acceleration, time of max angular velocity, duration time to contact. | > operated limb, maximum extension angle (p < 0.012)<br>> peak velocity (p > 0.033) |
| Pollard et al. 2015<br>ACLR | Intra-limb coupling angles; hip rotation – knee abd/add hip flex/ext – knee abd/add, hip rotation – ankle IN/EV, knee abd/add – knee flex/ext knee abd/add – ankle IN/EV, knee abd/add – knee rotation, knee flex/ext – knee rotation | > hip rotation – knee abd-add (p=0.04)<br>> hip flex/ext – knee abd-add (p=0.05)<br>> knee abd/add – knee flex/ext (p < 0.01)<br>> knee abd/add – knee rotn (p=0.03) |
| Gribbin et al. 2016<br>ACLR | Intra-limb joint couples; hip frontalknee frontal, hip frontal-knee sagittal, hip frontal-knee transverse, hip sagittal-knee frontal, hip sagittal-knee transverse, hip transverse-knee frontal | WALK<br>> hip frontal – knee frontal 24-32% gait cycle<br>(midstance) (cohen's d 11.7)<br>> hip frontal – knee frontal 49-53% gait cycle<br>(late stance) (cohen's d 4.5)<br>> hip frontal – knee transverse 51-58% gait cycle<br>(late stance) (cohen's d 7.3)<br>> hip sagittal – knee transverse 53-55% gait cycle (late stance) (cohen's d 7.69)<br>> hip sagittal – knee transverse 67-69% gait cycle<br>(swing) (cohen's d 35.85)<br>> hip transverse – knee frontal 27-29% gait cycle<br>(midstance) (cohen's d 14.9)<br>> hip transverse – knee frontal 45-47% gait cycle<br>(late stance) (cohen's d 12.0)<br><br>< hip sagittal – knee frontal 25-31% gait cycle<br>(midstance) (cohen's d -2.3)<br>< hip sagittal – knee transverse 25-30% gait cycle<br>(midstance) (cohen's d -2.69) |

Table A.1: All variables measured and significant findings (cont.)

| | | |
|---|---|---|
| | | RUN<br>> hip sagittal – knee transverse swing (not reported) |
| Kipp and Palmieri-Smith 2013<br>CAI | Ankle sagittal and frontal plane; touchdown, maximum, minimum<br>angle. Ankle sagittal and frontal peak moment. Sagittal and frontal principal components kinematic & kinetic time-series data | = touchdown angle, max/min angle sagittal/frontal plane<br>= peak moments, kinetic PC scores<br><br>> kinematic sagittal PC 3 (100ms pretouchdown)<br>> kinematic frontal PC 1 (through entire 300ms) |
| C. Brown, Bowser, and Simpson 2012<br>CAI | 3D kinematics ankle, knee, hip, trunk | PRE-INITIAL<br>> (FAI) trunk LF (p=0.006)<br><br>< (FAI) knee IR/ER (p=0.007)<br>< (coper) knee IR/ER (p=0.001) |
| | | < (MAI) hip flex (p=0.006)<br>< (FAI) hip flex (p=0.001)<br>< (coper) (p=0.006)<br>< coper anterior (p=0.007)<br><br>STANCE<br>< FAI knee IR/ER (p=0.003)<br>< MAI hip flex (p=0.003)<br>< FAI hip flex (p<0.001)<br>< Coper hip flex (p=0.001)<br>< MAI hip abd/add (p=0.003)<br><FAI lateral (p=0.003)<br>< MAI anterior (p=0.006)<br>< FAI anterior (p=0.005) |
| Hamacher, Hollander, and Zech 2016<br>CAI | Ankle sagittal and frontal plane angles | > injured ankle frontal plane 11-24% stance (p <<br>0.001)<br>> injured ankle frontal plane 77-83% swing<br>(p=0.005)<br>> injured ankle frontal plane 92-97% swing<br>(p=0.007)<br>> unaffected ankle frontal plane 66-69% swing (p=0.023) |
| Drewes et al. 2009<br>CAI | Rear-foot inversion-eversion, shank rotation. Coupling shank rotationrear-foot inversion-eversion | = No significant differences in DP measures between groups during walking or jogging (p > 0.05) |
| Herb et al. 2014<br>CAI | Joint coupling rear-foot and shank | WALK<br>< injured ankle rear-foot-shank late stance, toe off, early swing (not reported) |

Table A.1: All variables measured and significant findings (cont.)

| | | |
|---|---|---|
| Kulig, Joiner, and Chang 2015<br>Patellar tendon | Lower extremity contact angle, sagittal plane hip, knee and ankle | < injured limb ankle DF initial contact (p=0.01) |
| James, Dufek, and Bates 2000<br>Injury prone | Hip, knee, ankle; peak joint moments, time to peak moment values, impact impulse | > injured limb 100% jump height peak AJM (p < 0.05)<br>< injured limb 50% jump height time to peak AJM (p < 0.05) |
| Ferber, Davis, and Williams 2005<br>RRI | Intra-limb couple; rear-foot eversion/inversion - tibial internal/external rotation | = No significant differences in variability in joint coupling between groups |
| Mann et al. 2015<br>RRI | Strike Index, contact time, stride time, flight time, duty factor, stride length, stride frequency | = No significant differences in variability found between groups |
| Paquette, Milner, and Melcher 2016<br>RRI | Sagittal plane foot contact ankle | = No significant differences in variability of sagittal plane foot contact ankle between groups |
| Maclean, Emmerik, and Hamill 2010<br>RRI | Intra-limb couples; tibia transverse – calcaneus frontal, knee sagittal – rearfoot frontal, knee frontal – rear-foot frontal, knee transverse – rear-foot frontal | > Injured limb tibial transverse-calcaneus frontal early stance (p=0.004; ES=0.30) |
| Meardon, Hamill, and Derrick 2011<br>RRI | Stride time | = No significant differences in variability of stride time between groups at the beginning of the run |
| Miller et al. 2008<br>ITBS | Intra-limb couples; thigh frontal (Abd/Add) – tibia transverse (IR/ER), thigh frontal – foot transverse (IN/EV), tibia transverse – foot transverse (IN/EV), knee sagittal (Flex/Ext) – foot frontal (Abd/Add), knee frontal (Abd/Add) – foot transverse | > knee flex/ext – foot abd/add complete stride (p=0.02)<br>> knee flex/ext – foot abd/add swing (p=0.04)<br>> knee flex/ext – foot abd/add stance (p=0.02)<br><br>< tibial IR/ER – foot IN/EV heel strike (p=0.04) ^ discrete measure |
| Hein et al. 2012<br>ITBS | Intra-limb coupling; hip sagittal – knee sagittal, hip frontal – knee sagittal, knee sagittal-ankle sagittal, knee sagittal-ankle sagittal | = No significant differences in variability of dependent measures found between groups during stance phase, intervals of stance phase or continuously during |

Table A.1: All variables measured and significant findings (cont.)

| | | |
|---|---|---|
| Heiderscheit, Hamill, and Emmerik 2002<br>PFP | Stride duration and length. Withinlimb coupling; thigh transverse – leg transverse, thigh sagittal – leg sagittal, knee transverse – ankle transverse (IN), Knee sagittal – ankle transverse (IN), knee sagittal – ankle sagittal | > stride length preferred running speed (p=0.03,<br>ES 0.30) |
| Cunningham et al. 2014<br>PFP | Coupling angle variability; knee valgus/varus (KV) – ankle inversion/eversion (AI), knee valgus/varus – ankle plantar/dorsi flexion (AF), knee flexion/extension (KF) – ankle inversion/eversion, knee flexion/extension – ankle plantar/dorsi flexion, knee internal/external rotation (KR), knee internal/external – ankle plantar/dorsi flexion | > KF-AF Q1 (p=0.020; cohen's d 0.97)<br>> KR-AI Q2 (p=0.049; 0.80)<br>> KR-AF Q2 (p=0.038; 0.85)<br>> KV-AF Q4 (p=0.010; 1.09)<br>> KV-AF Q5 (p=0.008: 1.12)<br>> KV-AF stance (p=0.008: 1.21)<br>> KV-AI stride (p=0.031; 0.89) |
| Edwards et al. 2016<br>AGP | 3D kinematics ankle, knee, hip, L5-S1 and T12-L1. Peak net internal ankle, knee and hip joint moments. Peak ground reaction force (vertical, minima, posterior) | > T12-L1 right-left rotation (p < 0.05) |
| Chiu, Lu, and Chou 2010<br>THA | Inter-joint coordination variability; hip - knee, knee - ankle | > surgical limb hip-knee sagittal plane presurgery (p=0.019)<br>> surgical limb knee-ankle sagittal plane presurgery (p=0.008)<br>> surgical limb knee-ankle sagittal plane 6 weeks post-op (p=0.036) |

ACLR - anterior cruciate ligament reconstruction, CAI - chronic ankle instability, RRI - running related injury, ITBS - iliotibial band syndrome, PFP - patellofemoral pain, AGP - athletic groin pain, THA - total hip arthroplasty, > greater than, < less than, = no between group difference, ROM - range of motion, abd - abduction, add - adduction, flex - flexion, ext - extension, IN - inversion, EV - eversion, rotn - rotation, PC - principal component, max - maximum, min - minimum, FAI - functional ankle instability, MAI - mechanical ankle instability, IR - internal rotation, ER - external rotation, LF - lateral flexion, DP - deviation phase, DF - dorsi-flexion, AJM - ankle joint moment, ES - effect size, T12 - thoracic vertebrae 12, L1 - lumbar vertebrae 1.

Table A.2: Quality appraisal for variability systematic review

| | REPORTING | | | | | | | | | | | EXTERNAL | | | | INTERNAL BIAS | | | | | | | | CONFOUNDING | | | | | | | POWER | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Q1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 | | 11 | 12 | 13 | | 14 | 15 | 16 | 17 | 18 | 19 | 20 | | 21 | 22 | 23 | 24 | 25 | 26 | | 27 | |
| **Author** | | | | | | | | | | | | | | | | | | | | | | | | | | | | | | | | **TOTAL** |
| (van Uden et al. 2003) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 12 |
| (Cordeiro et al. 2015) | 1 | 1 | 0 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 8 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 12 |
| (Pollard et al. 2015) | 1 | 1 | 0 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 8 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 11 |
| (Gribbin et al. 2016) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 0 | 8 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 11 |
| (Herb et al 2014) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 0 | 8 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 11 |
| (Kipp and Palmieri-Smith 2013) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 12 |
| (Brown, Bowser, and Simpson 2012) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 13 |
| (Hamacher, Hollander, and Zech 2016) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 12 |
| (Drewes et al. 2009) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 0 | 8 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | | 0 | 1 | 2 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 11 |
| (Kulig, Joiner, and Chang 2015) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 12 |
| (James, Dufek, and Bates 2000) | 1 | 1 | 0 | 1 | 1 | 1 | 1 | 0 | 1 | 0 | 7 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 10 |
| (Ferber, Davis, and Williams 2005) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 1 | 0 | 1 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 14 |
| (Mann et al. 2015) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 13 |
| (Paquette, Milner, and Melcher 2016) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 13 |
| (Maclean, Emmerik, and Hamill 2010) | 1 | 1 | 0 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 8 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 12 |
| (Meardon, Hamill, and Derrick 2011) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 12 |
| (Miller et al. 2008) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 10 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 14 |
| (Hein et al. 2012) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 13 |
| (Heiderscheit, Hamill & Emmerik 2002) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 10 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 13 |
| (Cunningham et al. 2014) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 10 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 13 |
| (Edwards et al. 2016) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 10 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 14 |
| (Chiu, Lu, and Chou 2010) | 1 | 1 | 1 | 1 | 1 | 1 | 1 | 0 | 1 | 1 | 9 | 0 | 0 | 0 | 0 | 0 | 0 | 1 | 0 | 1 | 0 | 1 | 3 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 12 |
| | | | | | | | | | Mean | | 9 | | | | 0 | | | | | | | | 3 | | | | | | | 0 | | 12 |
| | | | | | | | | | Median | | 9 | | | | 0 | | | | | | | | 3 | | | | | | | 0 | | 12 |

# Appendix B

# Appendix B: The number of trials required when examining variability in athletes with athletic groin pain during a running cut

## B.1 Introduction

Movement variability is defined as the normal variations that occur in a motor performance across multiple repetitions of a task (Stergiou and Decker, 2011). Whilst traditionally within biomechanics movement variability has often been treated as noise or error (Racic, Pavic and Brownjohn, 2009; Taylor et al., 2015; Gore et al., 2016), it is possible that the variability has a functional role to play in injury (Hamill, Palmer and Van Emmerik, 2012; Baida et al., 2017). To date however there are a limited number of studies that have appropriately examined this concept, and their findings have not been consistent (Baida et al., 2017). For example, several authors have provided support for the association between injury and reduced variability (Hamill et al., 1999; Herb et al., 2014).

In contrast other authors have suggested that greater variability is associated with injury (Stergiou, Harbourne and Cavanaugh, 2006; Hamill, Palmer and Van Emmerik, 2012), while others have found no relationship between these two factors (Ferber et al., 2005; Drewes et al., 2009; Mann et al., 2015). A methodological consideration that may, at least in part, explain these contrasting findings is selecting an appropriate number of trials to adequately represent this variability (Baida et al., 2017). On the one hand, if too few trials are utilised then the total variability may not be adequately captured, potentially leading to erroneous findings. However, it may also not be possible to capture a large number of trials due to cost, time, fatigue and potentially causing further injury to participants.

The concept of acquiring a sufficient number of trials is not new with respect to measures of central tendency (Bates et al., 1983; Rodano and Squadrone, 2002; James et al., 2007; Racic, Pavic and Brownjohn, 2009; Gore et al., 2016) however, as noted in a recent systematic review by our research group (Baida et al., 2017), few studies investigating variability have included an experimental justification for the number of trials examined (Sangeux et al., 2016; Hafer and Boyer, 2017). One potential limitation of these studies however, is the sole use of statistical reliability as a criterion for how many trials are required when examining variability. For example, commonly the number of trials is determined based on the magnitude of variability staying within a given range (e.g. 10% relative precision) as the number of trials examined is increased (Sangeux et al., 2016; Hafer and Boyer, 2017). However, given that the number of trials required has been shown to be task dependent (Hafer and Boyer, 2017), and is likely specific to the population being examined, the number of trials required should be determined with regard to how many trials are required to maximise the ability to detect differences in the amount of variability between subjects of interest (e.g. in injured subjects with athletic groin pain vs. uninjured controls) (Baida et al., 2017). The method most readers will be familiar with is to use a linear measure of effect [e.g. Cohen' D

effect size (Cohen, 1988)], however it may also be beneficial to utilize a non linear machine learning technique such as support vector machines which can identify differences in groups that are not linearly separable.

Athletic groin pain (AGP) is a common injury in field sports that is characterised by repetitive running cuts (Werner et al., 2009; Orchard, Seward and Orchard, 2013; Thorborg et al., 2017). In light of the high occurrence rate (Waldén, Hägglund and Ekstrand, 2015), along with the chronic nature and high morbidity of the injury (Thorborg et al., 2017), the examination of movement variability in AGP is warranted. To date only one study has examined variability in AGP patients, however this study did not provide justification for the number of trials examined (n = 10) (Edwards, Brooke and Cook, 2017). This study will therefore aim to determine the number of trials required to achieve an appropriately high effect-size in the amount of variability in athletes with and without AGP during a running cut task.

*Note: the results of this study will be used in Chapter 8 to determine the affects of AGP on movement variability.*

## B.2 Methods

Twenty AGP subjects mean $\pm$ SD: age 23.3 $\pm$ 2.8 yrs., height 1.81 $\pm$ 0.06 m, weight 80.1 $\pm$ 11.1 kg) and twenty uninjured field sport athletes (mean $\pm$ SD: age 25.0 $\pm$ 4.9 yrs., height 1.80 $\pm$ 0.05 m, weight 81.23 $\pm$ 6.74 kg) were recruited into this study. Inclusion criteria for the AGP patients required that all subjects reported pain in the anterior hip and groin area during their chosen sporting activity and have symptoms of duration greater than 4 weeks (Falvey, King and Kinsella, 2015). Furthermore, all AGP subjects were required to have a stated intention of returning to the same level of pre-injury participation in competitive

multidirectional sport. Participants with AGP were subject to clinical history, examination and MRI imaging prior to entry into the study, by a Consultant Physician in Sport and Exercise Medicine as per Falvey, King and Kinsella, (2015). Subjects with hip joint arthrosis [grade 3 or higher on MRI (Li et al., 1988)], those who did not intend to return to pre-injury activity levels, and those with underlying medical conditions such as inflammatory arthropathy or infection were excluded. The Sports Surgery Clinic's ethics committee provided ethical approval (REF SSC0024) and all of the participants signed informed consent.

### B.2.1 Biomechanical Model

As per the Vicon Plug in Gait model (Vicon Motion Systems, Oxford, UK), 28 Reflective markers (14mm diameter) were placed at bony landmarks on the lower limbs, pelvis and trunk with an additional marker placed on the anterior aspect of the mid tibia and mid thigh bilaterally. The 'OSSCA' method as implemented in NEXUS 2 was utilised (Taylor et al., 2010) which includes the calculation of the hip joint centre and the functional knee axes using the symmetrical centre of rotation estimation (SCoRE) (Ehrig et al., 2006) and the symmetrical axis of rotation approach (SARA) (Ehrig et al., 2007), respectively. To estimate the hip joint centre, each participant completed a star arc pattern with his or her hip consisting of flexion-extension, abduction-adduction and internal-external rotation while standing on the contralateral limb. To calculate the knee rotation axes the subjects performed a squat movement pattern three times. Soft tissue artefact is then minimised during dynamic trials using the optimal common shape technique (OCST) (Taylor et al., 2005), where an optimum rigid marker configuration for each segment is formed to reduce the effects of skin elasticity.

### B.2.2 Biomechanical data capture

Three dimensional marker positions were tracked using 10 infrared cameras (Vicon - Bonita B10, UK), synchronized with two 40x60cm force platforms (AMTI - BP400600, USA) collecting ground reaction force data. Motion and force data were captured at a sampling frequency of 200 Hz and 1000 Hz, respectively. Prior to the experimental testing the subjects completed a standardized warm-up (Marshall et al. 2015) involving a three-minute jog at a self-selected pace, five body weight squats and two practice trials of the 110° cutting task. For the cut, participants ran as fast as possible toward a marker placed on the floor, made a single complete foot contact on the force plate, and performed a 110° cut with respect to the approach direction before running maximally to the finish (Figure. B.1). Both marker and force data were filtered using a fourth order Butterworth filter with a cut-off frequency of 15Hz (Kristianslund, Krosshaug and van den Bogert, 2013). Kinematic and kinetic calculations were performed in Nexus software (Vicon Motion Systems, Oxford, UK) and were subsequently exported to Matlab for statistical analysis. In addition to typical kinematic and kinetic measures generated by Nexus, vector coding was utilised as a measure of co-ordination (see methods section Chapter 8). Within-subject variability was assessed using standard deviation or the circular equivalent for co-ordination variability (Batschelet, 1981).

### B.2.3 Methods of determining the required number of trials

**Feature selection and generation** The first step involved calculating the cumulative variability (standard deviation) by adding one trial at a time from two to fifteen trials, thus creating fourteen cumulative values for each participant and each biomechanical feature/variable of interest. Fourteen standardized measures of effect between the AGP and uninjured group were then calculated for each

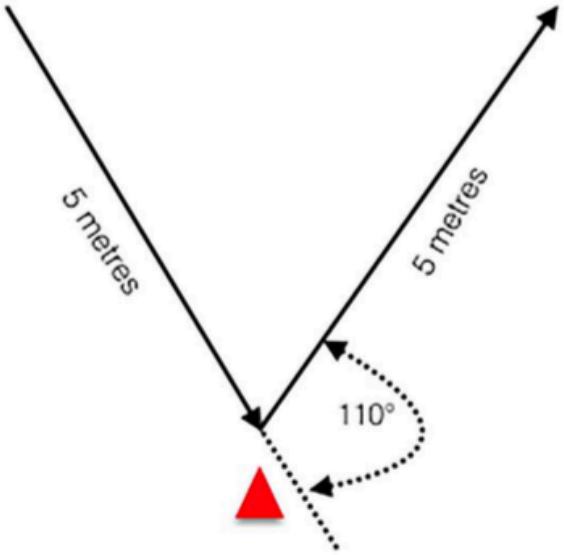


Figure B.1: Illustration of the 110° cut task

biomechanical feature representing the between-group difference calculated from the cumulative of two to fifteen trials. Given the positively skewed nature of variability at a population level, rank-biserial correlation (RBC) was utilized as a measure of effect size (Tomczak and Tomczak, 2014) (See appendix H for discussion on effect size choice). The score was presented as an absolute value between zero and one, with a value zero representing no effect. To retain only 'meaningful differences' in the calculation of the number of trials required, only stable features that demonstrated a consistent direction of difference (positive/negative effect) more than 85% of the time (12/14 cumulative trials) were retained. Furthermore, of these stable features, the associated scores were then sorted based on the absolute mean effect size and the features in the top quartile were selected for further analysis. This above process of feature generation and selection was conducted on a random 70% subset of the participants one hundred times to enhance the generalizability of the findings.

**Number of trials calculation** To decide the number of trials required at a whole body level, two approaches were utilised and examined visually.

**Mean effect size approach** For each of the one hundred random subsamples created in the feature generation and selection process, the generated RBC values were averaged across all retained features for each of the cumulative trials. This resulted in one hundred mean RBC curves (Figure B.2) representing the mean effect size calculated from the cumulative of two to fifteen trials. The mean of all one hundred mean RBC values was then calculated and utilized in the final decision of the number of trials required.

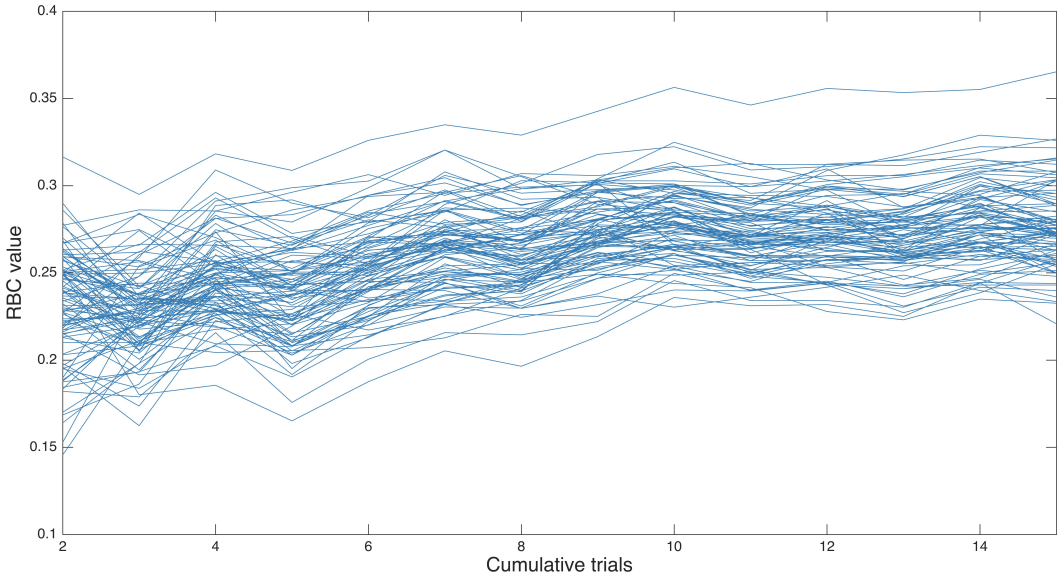


Figure B.2: Illustration of the one hundred mean RBC curves generated. Each curve represents the data from a single random 70% subsample. Within each curve, the value presented at each cumulative trial number represents mean RBC value calculated across all selected features at that cumulative trial number.

**Mean classification accuracy approach** The second approach involved calculating a mean five-fold cross-validated accuracy for each of the one hundred random subsample features selected in the feature generation and selection process. To do so, a support vector machine was utilized which maps the input features onto another feature space by a kernel function. The optimum hyperplane that separates the data in the mapped feature space can then be determined (Witten and Frank, 2005). After normalising the features to a zero mean and unit

variance, a support vector machine with a radial basis kernel was applied to the data using the 'fitcsvm' function as implemented in MATLAB.

## B.3 Results

The mean trials required differed depending on the statistical measure utilised and the metrics examined (Figure B.3 - B.5). For single joint kinematics measures, both the SVM cross-validated accuracy and mean RBC effect size reaches a peak value at seven trials and began to drop off thereafter , suggesting that seven trials are required (Figure B.3). For the co-ordination data, the RBC effect size did not vary greatly across the fifteen cumulative variables. Conversely, the SVM cross-validated accuracy reached its highest value at fifteen trials (Figure B.4). It is worth noting however, that there is a clear plateau from seven trials with only a marginal increase in SVM cross-validated accuracy thereafter (1.1%). For this reason seven trials were again chosen as the optimal number of trials required. Finally, for the kinetic measures, the mean RBC reached a maximum value at five trials while the SVM cross-validated accuracy fluctuated around 50% across all trials. Five trials were selected as the number needed for future research (Figure B.5).

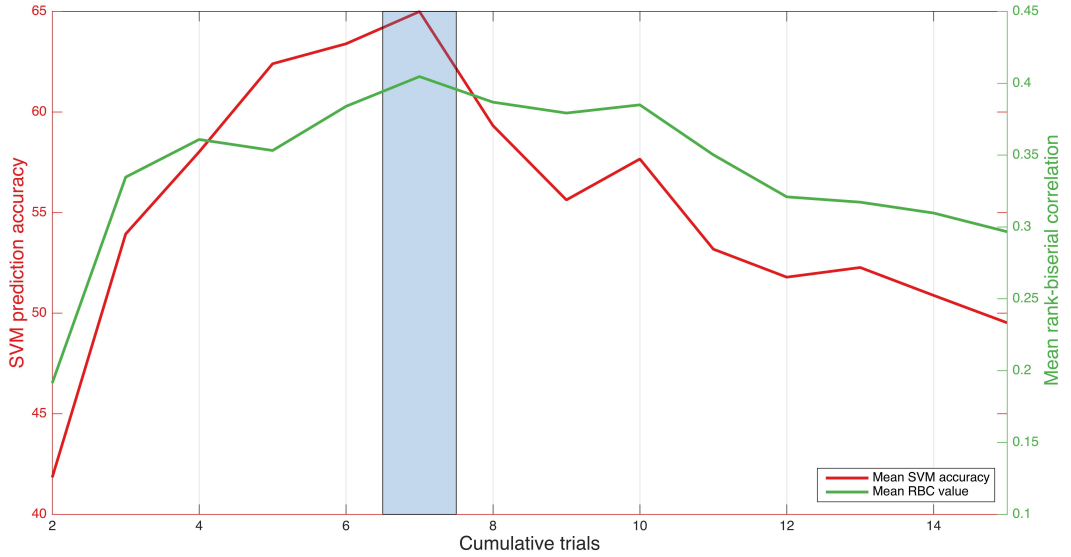


Figure B.3: Mean SVM accuracy and mean RBC for joint angles.Blue shaded area indicates the number of trials selected

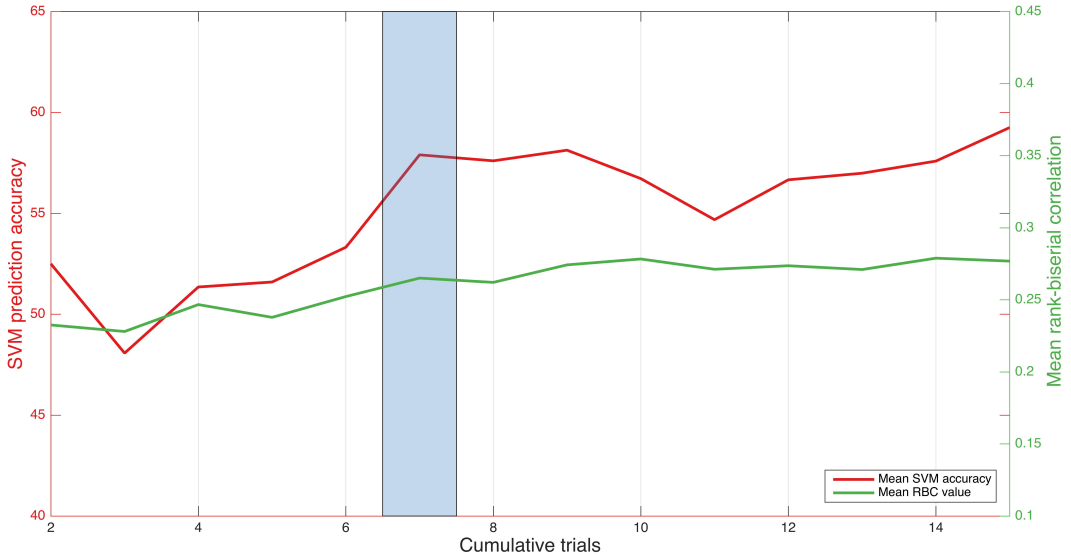


Figure B.4: Mean SVM accuracy and mean RBC for co-ordination. Blue shaded area indicates the number of trials selected

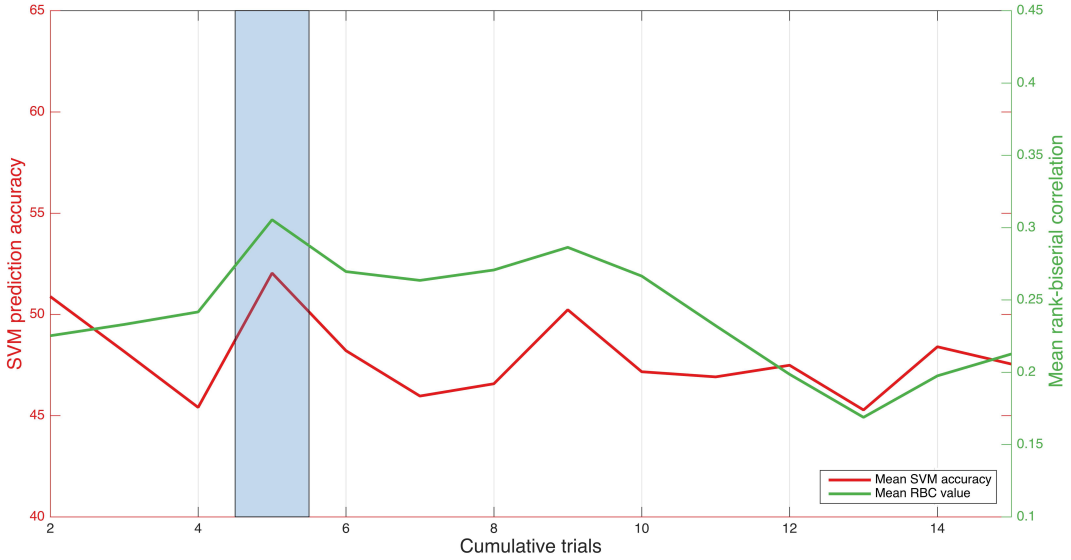


Figure B.5: Mean SVM accuracy and mean RBC for kinetics. Blue shaded area indicates the number of trials selected

## B.4 Discussion

Movement variability may be of importance with respect to injury (Baida et al., 2017). While it has been accepted that the number of trials obtained during an experiment can affect how representative measurements of central tendency will be (Bates et al., 1983; Hamill and McNiven, 1990; Rodano and Squadrone, 2002; James et al., 2007; Racic, Pavic and Brownjohn, 2009; Gore et al., 2016) to date few studies investigating variability have provided justification for the number of trials utilized (Sangeux et al., 2016; Hafer and Boyer, 2017). This methodological limitation may explain some of the contrasting findings in the literature with respect to variability and injury (Baida et al., 2017). The purpose of this study was therefore to investigate the number of trials required when comparing variability between those with and without AGP during a 110° cutting task. The findings from this study illustrate that seven trials are required when examining joint angle variability. Similarly, for co-ordination variability, while the highest mean SVM accuracy was recorded at fifteen trials, given the marginal increase in SVM accuracy (1.1%) and the relatively unchanging value, seven trials was again chosen as the optimal number required. This is similar to previous research that identified eight trials are required to achieve reliable co-ordination variability in healthy individuals during treadmill running (Hafer and Boyer, 2017). The findings from kinetic variability suggest five trials are required in the present study. While this study did not set out to explore whether variability is different between AGP patients and controls (which is examined fully in Chapter 8), it is worth noting, that the obtained SVM accuracies indicate that the model was generally no better than guessing group membership at random. This observation has two implications. Firstly, within the scope of this current experimental design, it would appear that kinetic variability may provide little information that is of relevance to AGP. Secondly, and perhaps more important to this current study, the

reported number of trials required for kinetic variability may be less robust than the angle and co-ordination variability. In contrast to these findings, joint angle variability demonstrated the greatest prediction accuracy (65%) and mean RBC value (0.40). This is in accordance with findings from a systematic review, which suggested that that kinematic variability might be the more sensitive to differences between groups than kinetic variability (Baida et al., 2017). In general, the reported effects sizes and prediction accuracy appear low, however the averaging of effect sizes and non-optimised feature selection process conducted in this study may be responsible for this. Future research is therefore warranted to compare AGP subjects to uninjured controls on a joint-by-joint level.

Interestingly within this study, both the joint angle variability and to a lesser extent kinetic variability, demonstrated a reduction in ability to distinguish between groups towards the latter trials in this experiment [e.g. trials 10 to 15 (Figures B.3 and B.5)]. This was quite unexpected as per classical test theory (Beckstead, 2013) the mean of a subject's observed results across multiple trials will asymptote to the true result as the trial number increases. Intuitively one might assume the same holds true for cumulative variability. Indeed this is the basic premise commonly utilised in single group statistical reliability methods, whereby a bandwidth of precision is typically centred around the mean/variability of all trials and the number of trials required is decided when the cumulative trials falls within this bandwidth (Hamill and McNiven, 1990; Rodano and Squadrone, 2002). While, this may be a valid assumption to make with respect to cyclic or non-demanding tasks, the exposure to multiple explosive discrete tasks such as the running cut may result in the accumulation of fatigue and/or the experience of pain, potentially rendering the variance of all trials unrepresentative. In fact is not uncommon for athletes with AGP to limit training intensity and/or volume in order to manage the pain associated with this condition (Hölmich and Thorborg, 2014). As such, the explanation of fatigue influencing AGP mechanics

is not without reason. However, it is interesting that this may have resulted in the masking of the between group differences present in early trials, since fatigue is generally associated with increased risk of injury (Moran and Marshall, 2006; Cortes et al., 2013). It is therefore plausible that fatigue is affecting the uninjured group and is resulting in variance of movement more similar to the AGP patients as the trials progress. It is not certain why co-ordination variability did not exhibit this same drop off in ability to distinguish between those with and without AGP, however this perhaps indicates that the variability in co-ordination (or lack of) is maintained despite the possible accumulation of fatigue and/or pain.

While previous research has examined the number of trials required using single-group statistical reliability for measures of central tendency (Bates et al., 1983; Rodano and Squadrone, 2002; James et al., 2007; Racic, Pavic and Brownjohn, 2009; Gore et al., 2016) and variability (Sangeux et al., 2016; Hafer and Boyer, 2017), to the best of the authors' knowledge, this is the first study to identify the number of trials required using between-group differences as the criterion measure. The benefit of this approach is that it avoids the potentially biased selection of bandwidths of precision and selects the number of trials required in an ecological setting with the ultimate goal of increasing the sensitivity of future studies to detect difference between groups of interest.

### B.4.1 Limitations

This study took a global whole body approach to determining the number of trials required with a focus on maximising the between-group differences. It is acknowledged however, that on an individual joint level, the number of trials may differ. In addition, while the approach taken in this study has a high ecological validity, the findings are not likely to be generalisable to other injuries and may

even be specific for the task examined. Generalisation of this studies' finding is not recommended.

## B.5 Conclusion

This study presented a novel but intuitively simple method of determining the number of trials required in an ecologically valid manner. The results demonstrated that seven trials are required when examining joint angle or co-ordination variability in AGP patients during a running cut task, while kinetic variability required five trials. Future research should perhaps adopt similar methodologies as presented within this study to determine the number of trials required between populations and tasks of interest. Further, the effects of AGP on movement biomechanics should be explored in detail using the number of trials indicated in this study.

# Appendix C

# Appendix C: Exercise Rehabilitation Program

The following appendix is a description of the rehabilitation program completed by the AGP patients at the Sports Surgery Clinic, Dublin, Ireland. Please note that this has been published in full elsewhere (King et al., 2018), and is presented here with minor edits. For a full description and justification of the exercise rehabilitation program please see (King et al., 2018). Each rehabilitation session was approximately 1 hour in length and took place with a physiotherapist every 12-16 days depending on subject availability. Two physiotherapists undertook all physical assessments and exercise prescription and reviewed their practice together on a bi-monthly basis to ensure continuity between them. The first rehabilitation session started on the same day as the initial diagnostic assessment. The subjects were taken through each of the level 1 rehabilitation exercise streams ( Table C.1 & C.2; Figure C.1) with the difficulty of the exercise selected progressed or regressed depending on the subjects ability to execute with appropriate technique. Exercise completion by the subject was captured on video using Dartfish software for all exercises and levels of the programme and the videos hosted online for the subject to review between sessions. The subjects were instructed to complete 3-4 sets of

6-8 reps of each exercise 4 times per week at their own training base. Where a weight was included (i.e. deadlift) the weight selected was modified to allow the subject reach the appropriate number of repetitions and progressed as strength improved. Subjects were advised that should any exercise reproduce their groin symptoms, they should review their videos and amend their technique to resolve and if unable to do so these exercises should be discontinued and their assigned physiotherapist contacted. The equipment required was a squat rack, a high and low box or chair and some cones. Each subject was given a printed handout of all the exercises and main cues for each, in addition to a link to video recordings of their exercises available online as well as a 'rehab tracker' which noted compliance as well as any issues with any of the exercises which could be assessed at the next review. Progression to level 2 was indicated once the subject had a negative crossover sign as determined by a lack of pain in the contralateral limb during resisted hip flexion (Brukner and Khan, 2012) and was usually achieved by the second session. Level 2 involved a progression of the level 1 exercises in each stream as competency allowed. Linear running sessions were also introduced 3 times per week, with at least one day rest in between, after completion of the level 1 session. These sessions commenced with linear running drills focusing on lumbopelvic control and posture, swing leg recovery and increased rate of force development (Table C.3 & C.4;Figure C.2 & C.3). The drills were carried out for 5-6 reps and 3-4 sets with a high emphasis on quality and intensity prior to commencing linear running programme A. Linear A running programme (Table C.5) was designed to complement the running drills while assessing the subjects' running load tolerance and suitability for progression. It started with low volume and low intensity, both of which increased at different points through the programme. If a subject had any increase in symptoms the morning after a running session, they were instructed to repeat the same running session when scheduled until they could tolerate it and then progress to the next session.

Progression to level 3 was indicated once the subject had completed the Linear A running programme symptom free and had symmetrical hip internal rotation at 90° hip flexion as well as pain free squeeze at 45°. In Level 3, the level 1 exercises were again progressed in difficulty with those streams that the athlete displayed full competency on being maintained in 1 session per week. The linear running was progressed to Linear B (Table C.6), which saw a reduction in volume but increase in intensity starting first from a rolling start but then progressing to a standing start. The linear running drills were maintained as a warm up to this session and were accompanied by multidirectional drills. The focus of the multidirectional drills was to improve segmental control, lateral rate of force development and improve agility prior to returning to sports specific movement (Tables C.7 & C.8). The drills were carried out for 5-6 reps and 3-4 sets with a high emphasis on quality and intensity.

Table C.1: Level 1 Exercise streams and reason for inclusion

| Stream | Reasons for inclusion |
|---|---|
| Hip Flexor | Stabiliser of anterior hip and key function in swing leg recovery during running and cutting |
| Lateral Hip Control | Improves femoracetabular dynamic control to minimise dynamic impingement |
| Abdominal | Improves oblique abdominal strength minimising excessive trunk rotation and pevlic tilt |
| Double Leg Squat | Improves hip & lumbopelvic strength and control, minimising dynamic impingement |
| Lateral Hip Strength | Improves hip abduction and external rotation strength |
| Deadlift | Improves lumbopelvic control and posterior chain strength |
| Lunge | Improves lumbopelvic control, quadriceps and hip strength |
| Plyometric | Improves rate of force development and single leg reactive strength |

Table C.2: Level 1 streams and progressions

| Stream | Progressions | | |
|---|---|---|---|
| **Hip Flexor** | Supine | Standing Supported | Free standing |
| **Lateral Hip Control** | Supported Hip Hitch | Free Standing Hip Hitch | Step Up |
| **Abdominal** | Crook Lying Leg Lift | Crook Lying Alternate Leg Drop | Pallof Kneeling Split Lunge |
| **Double Leg Squat** | High Goblet Squat | Low Goblet Squat | Front Squat |
| **Lateral Hip Strength** | Abduction/External rotation in mini squat | Abduction/External rotation in mini squat at wall | Banded Squat |
| **Deadlift** | Hip Hinge | ½ Rack Deadlift | Floor Deadlift |
| **Lunge** | Split Lunge | Overhead Split Lunge | Weighted Split Lunge |
| **Plyometric** | On Spot Hopping | Line Hopping | Cone Hopping |


Figure C.1: Intersegmental control and strength - Deadlift

Table C.3: Linear Running Drills and reason for inclusion

| Linear Drills | Reason for Inclusion |
|---|---|
| **Marching/Skipping** | Focus on maintaining neutral lumbopelvic position and trunk posture while maximising vertical ground reaction force production |
| **Barbell/Overhead Running** | Focus on maintaining neutral lumbopelvic position and minimising overstride and excessive trunk rotation |
| **Leg Change Drill** | Focus on stance leg stiffness and swing leg recovery |

Table C.4: Linear Running Drills Instruction

| Linear | Instructions |
|---|---|
| **Marching/Skipping** | March/Skip on the spot with arms locked overhead, maintaining lumbopelvic neutral and with aggressive ground contact |
| **Barbell/Ovehead Running** | Run with dowel overhead or barbell across shoulders focusing on tall running posture and keeping stick still |
| **Leg Change Drill** | in single leg stand focus on rapid leg change to drive alternating leg extension and swing leg recovery |
| Complete 5-6 reps of 3-4 sets | |
| Focus is entirely on quality of exercise execution | |



Table C.5: Linear A Running Programme

| Linear A | | | | | | |
|---|---|---|---|---|---|---|
| **Session** | **Distance (meteres)** | **Intensity** | **Recovery** | **Reps** | **Distance** | **Total Distance** |
| 1 | 400 | 50% | 1 min | 6 | 2400 | 2400 |
| 2 | 400 | 50% | 1 min | 8 | 3200 | 3200 |
| 3 | 400 | 50% | 1 min | 10 | 4000 | 4000 |
| 4 | 400 | 70% | 1 min | 10 | 4000 | 4000 |
| 5 | 400 | 85% | 1 min | 10 | 4000 | 4000 |
| 6 | 400 | 100% | 1 min | 10 | 4000 | 4000 |
| 100% intensity was the subjects self rated assessment of maximum effort at that distance | | | | | | |
| Subjects progressed to the next level of running if no increase in groin symptoms the next morning | | | | | | |

Table C.6: Linear B Running Programme

| Session | Distance | Intensity | Recovery | Reps | Distance | Total Distance | Starting Speed |
|---|---|---|---|---|---|---|---|
| Warm Up | 400 | 70% | 1 min | 4 | 1600 | 1600 | |
| 1 | 100 | 70% | 30 sec | 10 | 1000 | 2600 | Rolling start x 10m |
| 2 | 100 | 85% | 30 sec | 10 | 1000 | 2600 | Rolling start x 10m |
| 3 | 100 | 100% | 30 sec | 10 | 1000 | 2600 | Rolling start x 10m |
| 4 | 100 | 100% | 30 sec | 5 | 500 | 2350 | Rolling start x 10m |
| | 50 | 70% | 30 sec | 5 | 250 | | Standing start |
| 5 | 100 | 100% | 30 sec | 5 | 500 | 2350 | Rolling start x 10m |
| | 50 | 85% | 30 sec | 5 | 500 | | Standing start |
| 6 | 100 | 100% | 30 sec | 5 | 500 | 2350 | Rolling start x 10m |
| | 50 | 100% | 30 sec | 5 | 250 | | Standing start |
| 7 | 100 | 100% | 30 sec | 5 | 500 | 2600 | Rolling start x 10m |
| | 50 | 100% | 30 sec | 10 | 500 | | Standing start |
| Warm up was completed prior to commencing each session warm up and to re-establish good running pattern | | | | | | | |
| 100% intensity was the subjects self rated assessment of maximum effort at that distance | | | | | | | |
| Subjects progressed to the next level of running if no increase in groin symptoms the next morning | | | | | | | |

Figure C.2: Linear running mechanics - skipping

Table C.7: Multidirectional Running Drills and reason for inclusion

| Multidirectional Drills | Reason for Inclusion |
|---|---|
| **Lateral Shuffle** | To optimise frontal plane rate of force development and minimise loss of segmental control between the trunk and pelvis |
| **Zig Zag Running** | To optimise trunk and hip control and foot placement during side step |
| **180 Degree Cone Cutting** | To optimise rate of force development and push off during cutting |



Table C.8: Multidirectional Running Drills

| Multidirectional | Instructions |
|---|---|
| **Lateral Shuffle** | Side Shuffle between 2 cones 8 metres apart with arms locked overhead focusing on getting away from the cones as quickly as possible. Progressed to react to instruction or to shadow opponent while shuffling |
| **Zig Zag Cutting** | 5 cones in zig zag formation, 5 metres apart from each other. Run and cut as quickly as possible around the cones. Add holding a med ball for increase resistance and higher centre of mass |
| **180 Degree Cone cutting** | 5 cones in a semi circle, start in the middle and run at any cone and cut back straight to the starting point.Add holding a med ball for increase resistance and higher centre of mass |
| Start at 50% intensity and increase between sessions as long as symptom free during drill | |
| Focus is entirely on quality of exercise execution | |
| 3-4 sets of 5-6 reps | |

Figure C.3: Multidirectional Drill - Shuffle Drill

# Appendix D

# Appendix D: Additional Finding from Chapter 4

While the main aim of Chapter 4 was to explore the relationship between stiffness and AGP, for completeness of findings from Chapter 4 the following appendix provides the additional findings also calculated from the study. This allows the interested reader to explore further the results from this chapter without taking from the primary aim and message of the study.

Table D.1: Mean summed change in moments and vertical ground reaction force

| *Variable* | *AGP Pre* | *AGP Post* | *Uninjured* | *AGP Pre vs. Uninjured* | | *AGP Post vs. Uninjured* | |
|---|---|---|---|---|---|---|---|
| | **Mean ± SD** | **Mean ± SD** | **Mean ± SD** | **D** | **Sig** | **D** | **Sig** |
| **Peak GRF (N.Kg$^{-1}$)** | 25.83 ± 3.63 | 25.45 ± 3.63 | 29.22 ± 3.62 | 0.85 | 0.00 | 0.93 | 0.00 |
| **Ankle plantarflexor** | 32.83 ± 7.37 | 31.87 ± 7.55 | 39.95 ± 10.17 | 0.76 | 0.00 | 0.84 | 0.00 |
| **Hip abductor** | 9.14 ± 6.05 | 9.00 ± 6.05 | 9.69 ± 7.71 | 0.08 | 0.68 | 0.10 | 0.60 |
| **Knee external rotator** | 2.06 ± 1.42 | 2.08 ± 1.35 | 2.06 ± 2.35 | 0.00 | 0.98 | 0.01 | 0.97 |
| **Knee extensor** | 16.03 ± 6.18 | 13.94 ± 6.19 | 15.14 ± 7.5 | 0.13 | 0.49 | 0.18 | 0.36 |
| **Knee abductor** | 10.24 ± 7.26 | 10.4 ± 6.37 | 10.51 ± 7 | 0.04 | 0.85 | 0.02 | 0.94 |
| **Ankle evertor** | 1.57 ± 2.19 | 1.59 ± 1.88 | 1.35 ± 1.93 | 0.11 | 0.59 | 0.13 | 0.51 |
| **Ankle internal rotator** | 2.51 ± 2.46 | 2.69 ± 2.55 | 3.2 ± 2.67 | 0.27 | 0.16 | 0.20 | 0.31 |
| **Hip extensor** | 6.08 ± 6.75 | 6.54 ± 6.92 | 9.47 ± 9.28 | 0.42 | 0.04 | 0.36 | 0.07 |
| **Hip Internal rotator** | -1.07 ± 1.24 | -1.6 ± 1.6 | -1.52 ± 1.72 | 0.30 | 0.15 | 0.05 | 0.83 |

Moments measured as Moments ($Nm.Kg^{-1}$), Peak GRF = Peak vertical ground reaction force, D = Cohen's D effect size, Sig = significance (p). Arranged in order of Table (2), Chapter 4.

Table D.2: Mean summed range of motion

| *Variable* | *AGP Pre* | *AGP Post* | *Uninjured* | *AGP Pre vs. Uninjured* | | *AGP Post vs. Uninjured* | |
|---|---|---|---|---|---|---|---|
| | **Mean ± SD** | **Mean ± SD** | **Mean ± SD** | **D** | **Sig** | **D** | **Sig** |
| **COM displacement (m)** | 1.10 ± 0.18 | 1.06± 0.19 | 1.01± 0.17 | 0.47 | 0.01 | 0.24 | 0.21 |
| **Ankle dorsiflexion** | 32.21 ± 5.7 | 31.36 ± 6.44 | 33.00 ± 5.37 | 0.14 | 0.46 | 0.27 | 0.15 |
| **Hip adduction** | 2.89 ± 1.71 | 2.33 ± 1.53 | 2.27 ± 2.04 | 0.33 | 0.09 | 0.03 | 0.88 |
| **Knee Internal rotation** | 9.93 ± 6.51 | 9.18 ± 5.18 | 9.21 ± 6.19 | 0.11 | 0.56 | 0.01 | 0.98 |
| **Knee flexion** | 6.32 ± 3.5 | 5.66 ± 2.62 | 5.35 ± 3.14 | 0.29 | 0.13 | 0.11 | 0.58 |
| **Knee adduction** | 2.87 ± 2.6 | 2.8 ± 2.49 | 2.85 ± 2.08 | 0.01 | 0.96 | 0.02 | 0.90 |
| **Ankle inversion** | 1.37 ± 1.14 | 1.3 ± 1.02 | 1.5 ± 0.78 | 0.13 | 0.51 | 0.21 | 0.27 |
| **Ankle external rotation** | 4.3 ± 5.07 | 4.22 ± 3.94 | 2.37 ± 2.93 | 0.44 | 0.02 | 0.51 | 0.01 |
| **Hip flexion** | 1.41 ± 1.17 | 1.5 ± 1.32 | 1.5 ± 1.51 | 0.07 | 0.73 | 0.00 | 0.98 |
| **Hip external rotation** | 2.44 ± 2.73 | 2.36 ± 2.49 | 2.98 ± 2.86 | 0.19 | 0.36 | 0.23 | 0.27 |

Range of motion measured as degrees (°), COM Displacement = centre of mass displacement, D = Cohen's D effect size, Sig = significance (p). Arranged in order of Table (2), Chapter 4.

Table D.3: Peak eccentric moments

| *Variable* | *AGP Pre* | *AGP Post* | *Uninjured* | *AGP Pre vs. Uninjured* | | *AGP Post vs. Uninjured* | |
|---|---|---|---|---|---|---|---|
| | **Mean ± SD** | **Mean ± SD** | **Mean ± SD** | **D** | **Sig** | **D** | **Sig** |
| **Ankle plantarflexor** | 33.18 ± 7.25 | 31.99 ± 7.05 | 40.23 ± 10.05 | 0.76 | 0.00 | 0.88 | 0.00 |
| **Hip abductor** | 12.38 ± 5.57 | 12.76 ± 5.43 | 15.16 ± 6.32 | 0.46 | 0.02 | 0.40 | 0.04 |
| **Knee external rotator** | 2.55 ± 1.53 | 2.67 ± 1.57 | 2.49 ± 1.92 | 0.03 | 0.86 | 0.10 | 0.59 |
| **Knee extensor** | 17.76 ± 5.41 | 15.06 ± 5.93 | 17.57 ± 7.56 | 0.03 | 0.88 | 0.37 | 0.05 |
| **Knee abductor** | 9.09 ± 5.39 | 9.26 ± 4.99 | 8.87 ± 6.14 | 0.04 | 0.84 | 0.07 | 0.71 |
| **Ankle evertor** | 3.22 ± 1.89 | 3.45 ± 2.09 | 3.55 ± 2.31 | 0.16 | 0.42 | 0.04 | 0.82 |
| **Ankle internal rotator** | 2.52 ± 2.35 | 2.35 ± 2.29 | 2.91 ± 2.7 | 0.16 | 0.42 | 0.23 | 0.24 |
| **Hip extensor** | 10.16 ± 6.46 | 11.67 ± 7.85 | 15.76 ± 8.87 | 0.70 | 0.00 | 0.48 | 0.02 |
| **Hip internal rotator** | 1.09 ± 0.66 | 1.26 ± 0.93 | 1.33 ± 0.74 | 0.34 | 0.10 | 0.08 | 0.69 |

Moments measured as Moments ($Nm.Kg^{-1}$), D = Cohen's D effect size, Sig = significance (p). Arranged in order of Table (2), Chapter 4.

# Appendix E

# Appendix E: Further information on stiffness calculation in Chapter 4

Derived from Hooke' s Law, stiffness is calculated as the change in force divided by the resulting displacement. This can graphically be visualised as the slope of a force vs. displacement graph (Figure E.1).

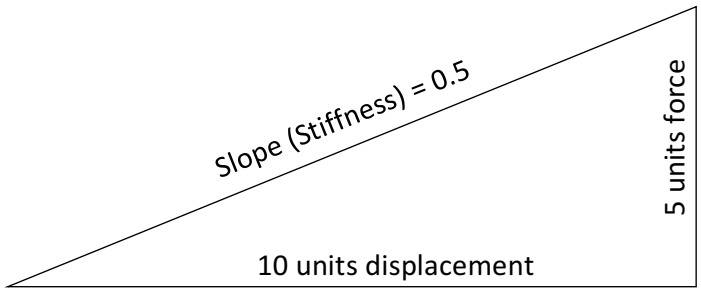


Figure E.1: Slope of a force vs. displacement graph

As noted in section 2.4 the strict physics definition of stiffness cannot be applied to measure joint stiffness in human movement. This was described in detail by Latash and Zatsiorsky, (1993), who highlighted that stiffness measures assume a passive body in which there exists a one to one relationship between force and body geometry with a resistance to deformation that is independent of time, length, or velocity. Clearly this is not the case for joint stiffness which is controlled

in part by muscles actively generating joint moments. Latash and Zatsiorsky, (1993) proposed the use of the term 'quasi-stiffness' to describe the simplified model of stiffness examined in human movement. While it is important to note the distinction between true mechanical stiffness and quasi-stiffness (explored within this thesis), for simplicity the term 'stiffness' was used throughout.

At a joint level within this thesis, stiffness was calculated only during phases where the joint was absorbing power (see methods section Chapter 4 for justification). To account for the often-intermittent nature and varying duration of stiffness during these phases, it was required to weight the stiffness calculated within each identified phase of interest by the magnitude of change in angle. Given a series of slopes, ($m_1, m_2, m_3, ...m_n$) calculated within each phase and defined by the equation of a slope:

$$m_i = \frac{Y_{2_i} - Y_{1_i}}{X_{2_i} - X_{1_i}} \tag{E.1}$$

The mean slope cannot be calculated by taking a simple arithmetic mean of the slopes:

$$\bar{m} \neq \frac{1}{n} \sum_{i=1}^{n} m_i \tag{E.2}$$

Rather, the mean stiffness must be calculated by either firstly summing all the changes in x and y for each phase and dividing their sums:

$$\bar{m} = \frac{\sum_{i=1}^{n} \Delta y_i}{\sum_{i=1}^{n} \Delta x_i} \tag{E.3}$$

Or as conducted in Chapter 4, by calculating the slope of the line within each phase, weighting each slope by it's corresponding change in x and then subsequently dividing the summation of all changes in x:

$$\bar{m} = \frac{[\frac{1}{n}\sum_{i=1}^{n} m_i * \Delta x_i]}{\sum_{i=1}^{n} \Delta x_i} \tag{E.4}$$

Within Chapter 4 joint stiffness was calculated as the change in moment divided by the change in joint angle. It is worth noting that within the 'plug in gait' model utilised in this study, joint angles are defined with respect to the proximal segment, while joint moments are calculated relative to the distal segment. This may introduce an error into the calculation of joint stiffness that should be considered by the reader.

# Appendix F

# Appendix F: Biomechanical Waveforms from Chapter 7.

While the main aim of Chapter 7 was to explore the relationship between change in biomechanics and change in a HAGOS-FSR score, the following angle graphs are provided to give the interested reader additional visual context of the running cut task.

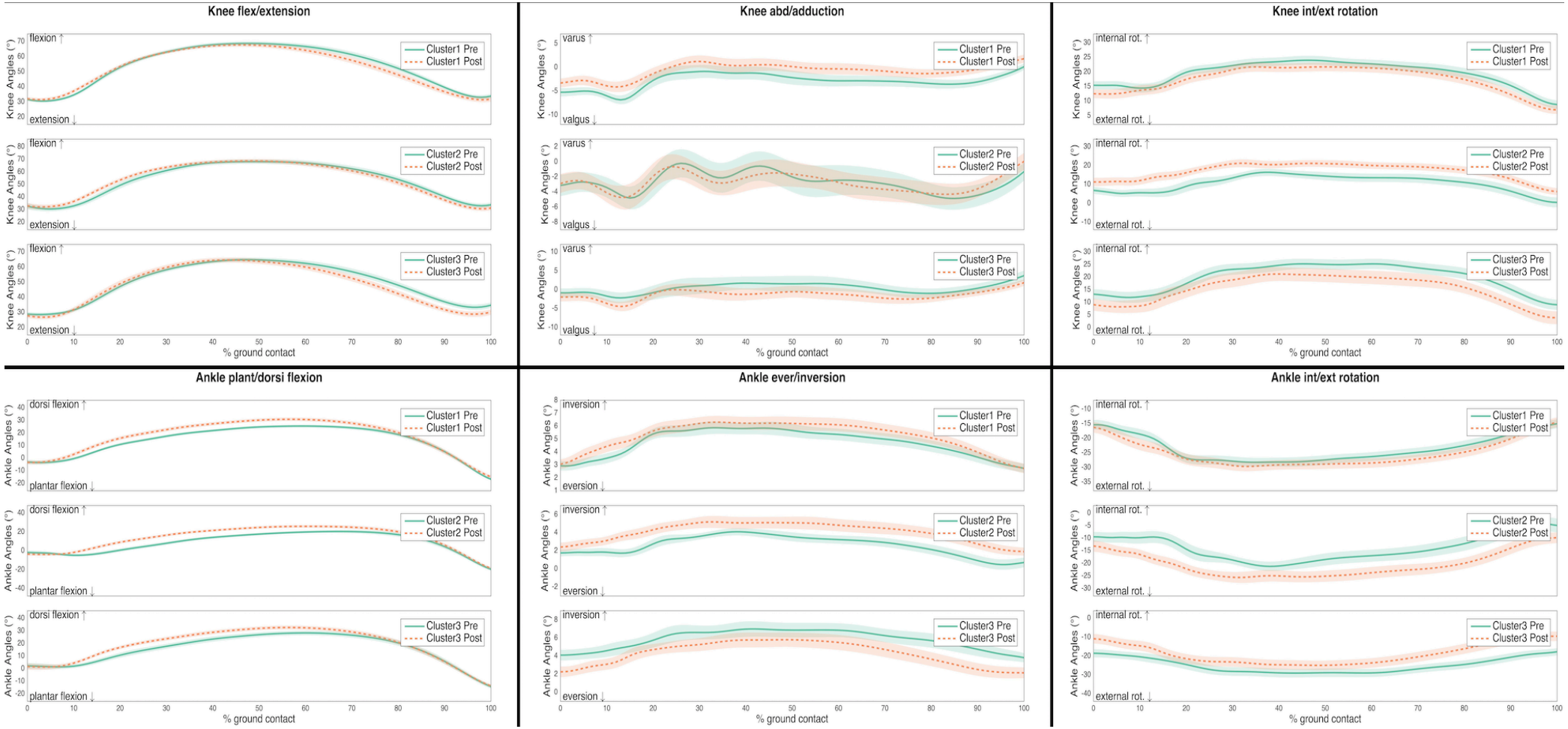


Figure F.1: Knee and ankle angles. Shaded region indicates standard error

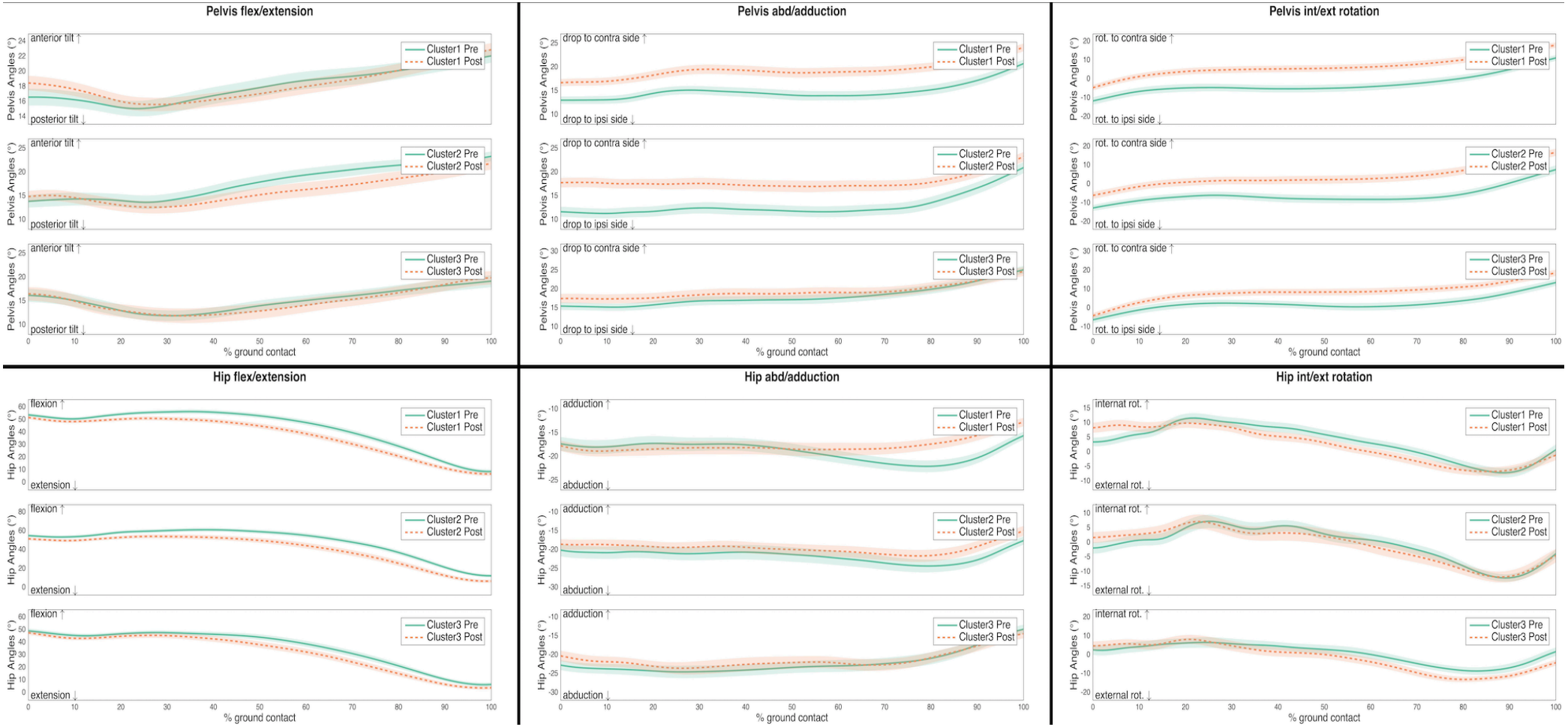


Figure F.2: Pelvis and hip angles. Shaded region indicates standard error

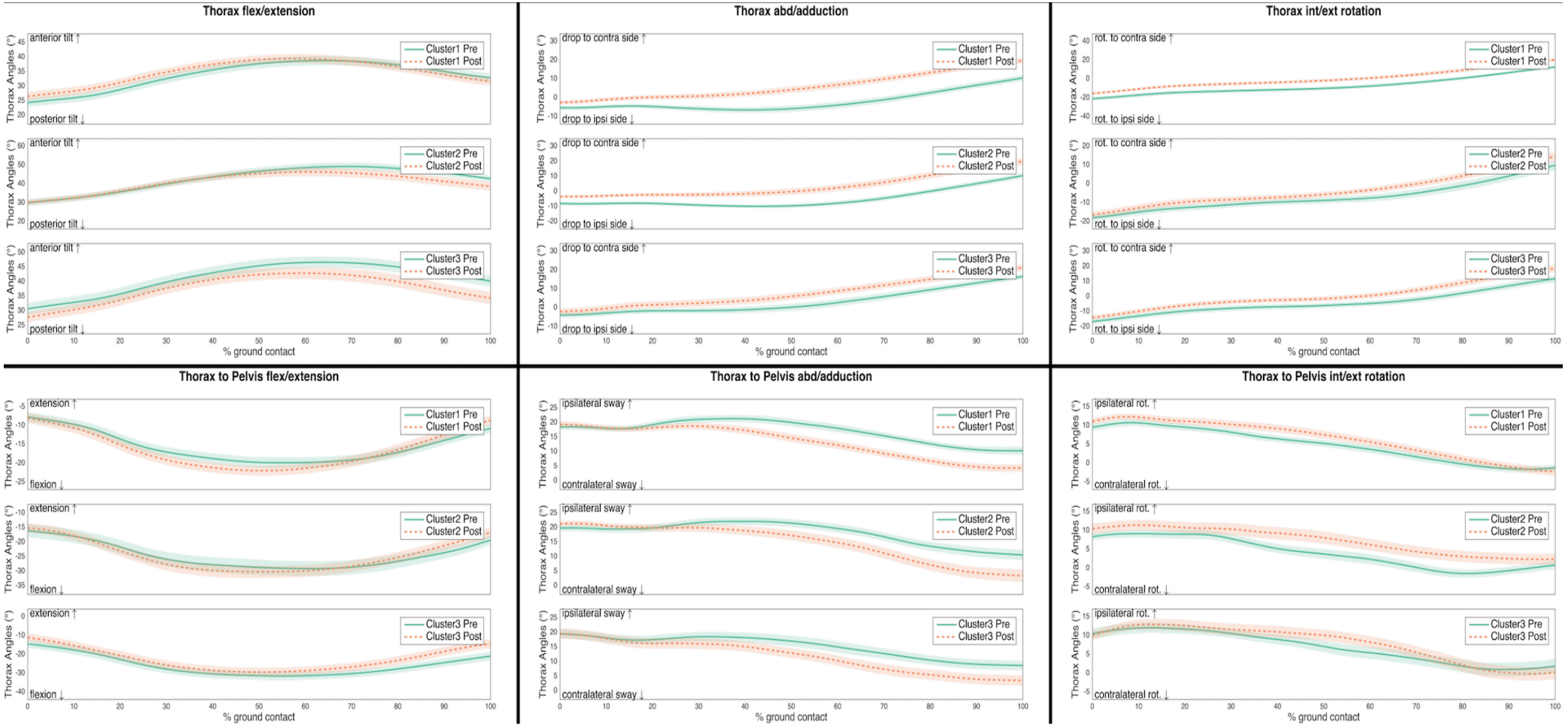


Figure F.3: Thorax and thorax to pelvis angles. Shaded region indicates standard error

# Appendix G

# Appendix G: Justification for choosing Vector Coding over Continuous Relative Phases for the Co-ordination calculation in Appendix B & chapter 8

Within the co-ordination literature, there are two main forms of co-ordination analysis, namely vector coding and continuous relative phases. To date only one study has directly compared both these measures of co-ordination for the examination of variability (Miller et al., 2010). The authors concluded that while both measures appear to provide valid metrics for assessing variability, within their study both measures did not convey the same information on variability at all times. In light of this, the current appendix section will illustrate the exploration of continuous relative phases as a measure of co-ordination and explain in brief the reason for choosing vector coding within this thesis.

**Continuous relative phase** CRP is typically derived from the position - velocity phase planes of two predominantly sinusoidal oscillators. Once the arctangent phase angle is calculated from position vs. normalised velocity plot of a single joint. The phase angle from one joint is subtracted from the phase angle of another joint of interest for each time point in the cycle (Hamill, Palmer and Van Emmerik, 2012).

$$CRP_i = \varphi 1_i - \varphi 2_i \tag{G.1}$$

Where $\varphi 1$ and $\varphi 2$ are the normalized phase angles for the joint 1 and 2, respectively, calculated at each time point $i$.

The biggest disadvantage of CRP however is its assumption that the two oscillating segments under scrutiny are of a one-to-one frequency ratio and they exhibit a sinusoidal time history (Hamill, Haddad and McDermott, 2000). While there are various normalisation techniques available for both frequency and amplitude (Burgess-Limerick, Abernethy and Neal, 1993; Hamill et al., 1999), a review of CRP conducted in 2014 illustrated that the normalisation techniques are also subject to a number of limitations (Lamb and Stöckl, 2014). The authors therefor proposed the use of a Hilbert transform. The Hilbert transform creates an analytic signal from non-sinusoidal signals, thereby removing frequency artefacts and making it appropriate for studying coordination in human movement (Rosenblum et al., 2003).

To assess the implementation of the method proposed by (Lamb and Stöckl, 2014), two sine waves were created with a known phase shift of 20° (Figure G.1.A). Both the traditional normalisation approach as outlined in (Hamill et al., 1999) that scales data to the unit phase space and a Hilbert transform were applied to the two signals to calculate the phase shift.

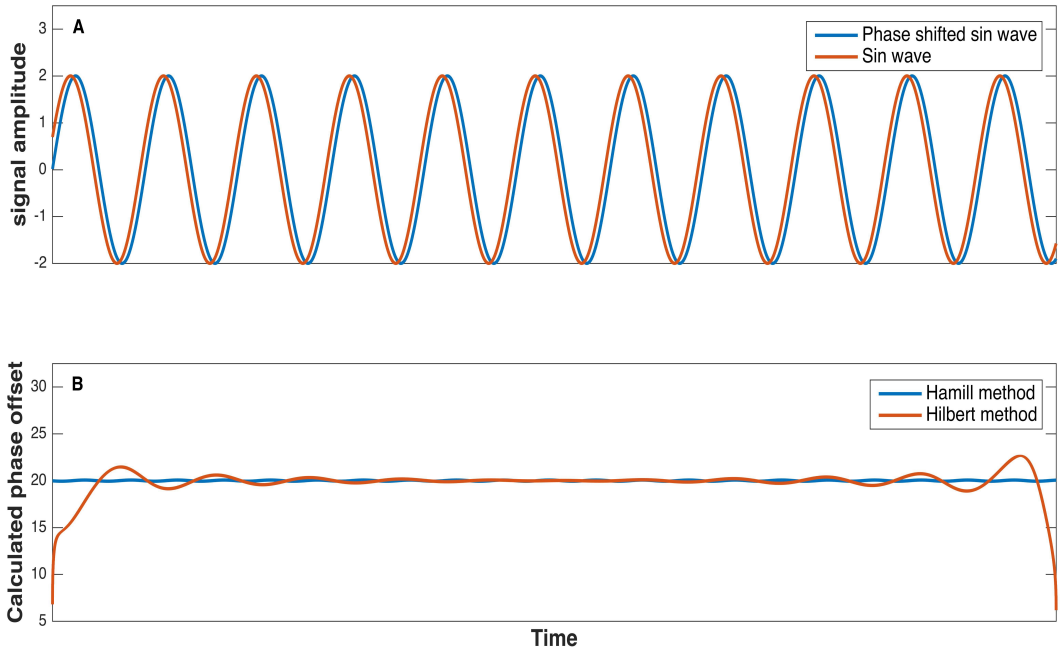


Figure G.1: Illustration of two different methods of calculating continuous relative phase. A). Plot of sine wave alongside the same sine wave phase shifted by 20°. B) Illustration of the results from the Hilbert method and method as presented by (Hamill et al., 1999)

**Results and Conclusion** As can be seen in Figure G.1.B, both methods provide an accurate measure of the phase shift. With the Hilbert transform method however, at both the beginning and end of the waveform there are large oscillations in the signal. This is a known distortion caused by the Hilbert transform and the first and last cycle are typically removed to avoid its artefact (Rosenblum et al., 2003). The oscillations observed in the generated signal of this appendix suggest that more than one cycle needs to be removed to obtain an accurate measure. Regardless, the distortions created during the Hilbert transform mean that this approach is not appropriate of use with discrete tasks as examined within this current thesis. For this reason, vector coding was chosen as measure of coordination within appendix B & chapter 8.

# Appendix H

# Appendix H: Justification for use of effect sizes and the choice of effect size utilised in Appendix B & Chapter 8.

While null hypothesis significance testing remains the standard method to evaluate a research question, a recent statement from the American statistical association noted some of the weaknesses associated with using p values alone to evaluate a hypothesis (Wasserstein and Lazar, 2016). In particular, the statement noted that a p-value, or statistical significance, does not measure the size of an effect or the importance of a result by itself and that a p-value does not provide a good measure of evidence regarding a model or hypothesis. For this reason, the use of effect sizes has also been utilised throughout this thesis (Sullivan and Feinn, 2012). An effects size is the magnitude of the difference between groups of interest. For example, the most simple form of effect is the absolute effect size which is the magnitude of difference in group means. Another form of effect size is the standardised effect size. The benefit of using a standardised effect size is that it

takes into account the variance in the data and also allows comparison across different metrics and studies (Sullivan and Feinn, 2012). The most common form of standardised effect size when comparing the mean difference between two groups is Cohen's D effect size (Ivarsson et al., 2013). The formula for calculating Cohen's D effect size is as follows:

$$D = \frac{M_1 - M_2}{\frac{\sqrt{(SD_1{}^2 + SD_2{}^2)}}{2}} \tag{H.1}$$

Where $M_1$ and $M_2$ are the group means and $SD_1$ and $SD_2$ are the groups' standard deviations (Ferguson, 2009) . As noted by Ivarsson et al. (2013) however, Cohen's D requires data that are reasonably normally distributed. As such, when using non- parametric statistics, non- parametric effect sizes should also be utilised. Despite this, the use of non- parametric effect siizes still appears very rare (Tomczak and Tomczak, 2014). In the absence of a common effect size utilised with non- parametric data, the following section will outline some of the non-parametric effect sizes explored and the process of selecting an appropriate effect size for use within appendix B & chapter 8 of this thesis.

**Methods** The cumulative variability from 2 to 15 trials was calculated and the between group effect size was calculated at each iteration (see chapter 8 for further details). Four effect sizes were plotted along with the mean accuracy from five-fold cross-validated non-linear support vector machine (Witten and Frank, 2005) for visual inspection (Figure H.1). In addition to Cohen's D, the following three nonparametric effect size metrics were chosen to explore:

**Cohen's $U_3$** Cohen's U3 is the measure of overlap between two distributions. It is calculated as follows:

$$U_3 = \frac{n_{1<median(2)} + 0.5 * n_{1=median(2)}}{n_1} \tag{H.2}$$

Where $n_{1<median(2)}$ is the number of elements in group 1 that are exceeded by the median value of group 2, $n_{1=median(2)}$ is the number of elements in group 1 that are equal to median value of group 2 and $n_1$ is the total number of elements in group 1.

**Area under the receiver operator curve (AUROC)** The receiver operator curve is a plot of the sensitivity (true positives) vs. 1 - the specificity (false positives) of a test. The area under this curve is calculated, with an ideal test having an AUROC value of 1, whereas a random guess would have an AUROC of 0.5 (Bewick, Cheek and Ball, 2004). The formula for AUCROC is as follows:

$$AUROC = \frac{1}{2}\sum(F_{i+1} - F_i)(T_{i+1} - T_i) \tag{H.3}$$

Where F is the number of false positives, and T is the number of true positives.

**Rank Biserial correlation** Rank Biserial correlation (RBC) is a correlation between a ranking and a dichotomy. Similar to Pearson's product-moment correlation, values range from -1 to +1 with a value of 0 indicating no effect.

$$RBC = \frac{2(\bar{R}_1 - \bar{R}_2)}{n_1 + n_2} \tag{H.4}$$

Where $\bar{R}_1$ and $\bar{R}_2$ are the mean ranks for group 1 and 2 respectively while $n_1$ and $n_2$ are the group's sample sizes (Tomczak and Tomczak, 2014).

**Findings and Conclusion** As can be seen from figure H.1, there was relatively little difference between the metrics examined. Interestingly, despite the fact the Cohen's D is only suitable for normally distributed data (Ivarsson et al., 2013), it

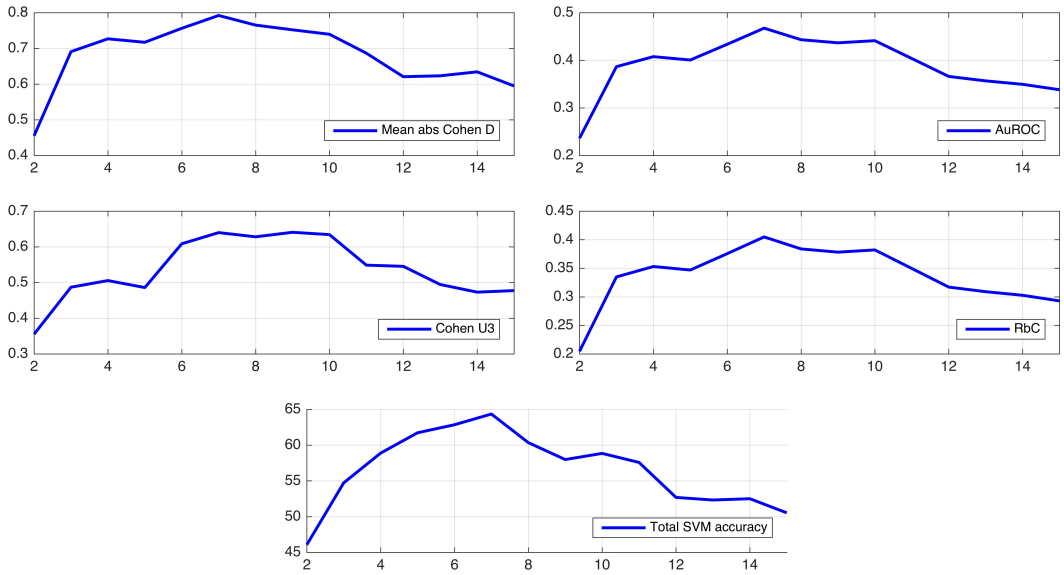


Figure H.1: Visual Inspection of different standardised measures of effect

did not appear to be affected by the distribution of the variability data, at least for the variables examined here. Of the effect size measures examined, Cohen's $U_3$ was the most dissimilar and did not exhibit the clear peak at seven trials present in every other measure. For this reason, rank biserial correlation was chosen as a nonparametric measure of effect within this thesis with support vector machine accuracy also utilised as secondary measure within chapter 8.

# Appendix I

# Appendix I: Overview of Analysis of Characterising Phases and Statistical Parametric Mapping

**Analysis of Characterizing Phases** Analysis of Characterizing Phases (ACP) is a technique, which examines key phases within a biomechanical waveform as identified by a VARIMAX rotated principle component analysis (Richter, O'Connor, et al., 2014a). Principle component analysis is a dimensionality reduction tool, which orthogonally transforms the original data onto new planes to best describe variance in the data (Robertson et al., 2013). Each transformation creates a principle component with an assigned Eigen value that represents the influence of the principal component on the original data set. The first principle component is the axes or plane, which explains the highest amount of variance in the original data with subsequent principle components explaining declining amounts of variance. ACP retains the number of principle components that explains 99% of the variance present within the biomechanical waveform (Richter et al., 2014).

Once the principle components have been extracted from the original data, each principle component's loading vector is examined and the key phase is then identified as follows:

For each principle component, the maximum loading is identified. The last value differing in sign before and after the absolute maximum then defines the start and end of the key phase, respectively. Each key phase is then separated into thresholds (e.g. 100%, 95% and 90% of the principal component peak) with the pattern-characterizing potential of the principal component reducing as the phase extends from principal component peak. The choice of what percentage of the peak principal component to retain has differed amongst publications (Richter, O'Connor, et al., 2014a; Marshall et al., 2015; Franklyn-Miller et al., 2017). Within this thesis however the following process was taken similar to previous research (Richter, O'Connor, et al., 2014a).

After identifying the first segment of a key phase (e.g. 95-100%), a phase score was generated representing, for example, the mean value within the phase. This phase is then tested and if a defined criteria (e.g. statistical significance or clear substantial difference (Batterham and Hopkins, 2006) between groups) is met then the phase is extended and retested. This process is terminated when the defined criteria are no longer met, and/or the start point/ the end point of the key phase is reached. The final phase retained is the 'characterising phase' examined.

**Statistical Parametric Mapping** Statistical parametric mapping was originally developed in the field of neuroimaging for the analysis of cerebral blood flow from 3D functional magnetic resonance imaging (Friston et al., 1994). As a technique, it requires data which is smooth and bounded. Biomechanical data typically adheres to these requirements since biological tissue exhibits viscoelastic properties, biomechanical data is sampled above Nyquist frequency, and data can be registered to key events (e.g. touch down and toe off) (Pataky, 2010). The

advantage of statistical parametric mapping is that it allows for the statistical testing of all data points within a waveform while controlling for the problem of multiple comparisons or 'family wise error'.

One standard method of dealing with multiple comparisons is to use the Bonferroni correction. With a Bonferroni correction, one adjusts the alpha level by dividing it by the number of observations made, where alpha is the false positive rate you are prepared to accept. For example, using a standard alpha level of 0.05 and a data set containing 101 data points (normalize to 100% of the action examined), the Bonferroni corrected p value would be 0.05 / 101 = 0.0004. This resulting Bonferroni corrected alpha level or critical threshold is likely to be too conservative for biomechanical data since neighbouring data points in a waveform are not independent, that is they may be highly correlated (Pataky, 2012).

Within statistical parametric mapping, for each time point in the registered waveform, a statistic value is calculated thereby approximating a continuous statistic trajectory (e.g. t values). Random field theory (Adler, 1981) then provides the mathematical foundation for conducting topological statistical inference and is charged with solving the problem of multiple comparisons in spatially correlated data. Random field theory first estimates the smoothness (spatial correlation) of the curve. This smoothness value is then used to calculate the expected Euler characteristic at different thresholds. This allows the identification of the critical threshold at which we would expect 5% of random data arising under the null hypothesis to contain at least one area above the threshold (Pataky, Vanrenterghem and Robinson, 2015).

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