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## Is There A Relationship Between Hip Structure, Hip Muscle Strength, and Lower Extremity Frontal Plane Kinematics During Treadmill Running?

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IS THERE A RELATIONSHIP BETWEEN HIP STRUCTURE, HIP MUSCLE STRENGTH, AND  
LOWER EXTREMITY FRONTAL PLANE KINEMATICS DURING TREADMILL RUNNING?

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THESIS

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A thesis submitted in partial fulfillment of the  
requirements for the degree of  
Master of Science in the College of Education  
at the University of Kentucky

By

Michael William Robinson Baggaley

Lexington, Kentucky

Co-Directors: Dr. Michael Pohl, Assistant Professor of Kinesiology  
and Dr. Jody Clasey, Professor of Kinesiology

Lexington, Kentucky

2014

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## ABSTRACT OF THESIS

### IS THERE A RELATIONSHIP BETWEEN HIP STRUCTURE, HIP MUSCLE STRENGTH, AND LOWER EXTREMITY FRONTAL PLANE KINEMATICS DURING TREADMILL RUNNING?

**INTRODUCTION:** Excessive hip adduction (HADD) has been associated with a number of lower extremity overuse injuries, and it has been suggested that it may be the result of reduced strength of the hip abduction musculature. Hip structure has been postulated to influence both hip abduction (HABD) strength and HADD. The purpose of this study was to investigate the relationship between hip structure, HABD strength, and frontal plane kinematics during running. **METHODS:** Peak isometric HABD strength, lower extremity kinematics, femoral neck-shaft angle (NSA), and pelvis width-femur length (pw-fl) ratio were recorded for 25 female subjects. Pearson correlations ( $P < .05$ ) were performed between variables. **RESULTS:** A fair relationship was observed between femoral NSA and HABD strength ( $r = -.472$   $P = .017$ ) where an increased NSA was associated with reduced HABD strength. No relationship was observed between HABD strength and frontal plane kinematics or between NSA/pw-fl and frontal plane kinematics. **CONCLUSION:** Alterations in the femoral NSA have the ability to influence peak isometric hip abduction strength. However, alterations in strength did not result in changes in lower extremity kinematics. Structural deviations at the hip do not appear to influence hip kinematics during running.

**KEYWORDS:** Running, Biomechanics, Hip, Strength, Structure

Michael William Robinson Baggaley  
August 5, 2014

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LOWER EXTREMITY FRONTAL PLANE KINEMATICS DURING TREADMILL RUNNING?

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I would like to dedicate this project to my advisor Dr. Michael Pohl for providing me with the opportunity to study at the University of Kentucky and for providing endless support and feedback during my time here. I would also like to dedicate this project to my parents Audrey Robinson and Carman Baggaley and my brother Daniel Baggaley. Without their continued love and support I would not be the person I am today.

## Acknowledgements

The following thesis, while an individual work, benefited from the insights and directions of several people. First, my Thesis Chair, Dr. Michael Pohl, provided critical feedback and guidance throughout the process, and this project would not have been a success without this assistance. Next, I would like to thank the other members of my committee: Dr. Robert Shapiro, Dr. Brian Noehren, and Dr. Jody Clasey. Each individual provided insight and feedback that greatly improved the quality of the final product. Lastly, I would like to thank Steven Capehart and Doug Long for all of their assistance with collecting data. This project could not have been completed without their help.

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## Chapter One: Literature Review

### Running Injuries

Running is one of the most popular recreational activities in the United States with 15 million Americans completing a race in 2012. Despite being one of the most popular recreational activities, some studies have reported rates of injury among runners up to 79% (1, 2). The knee was the most common site of injury (42%) with patellofemoral pain syndrome (PFPS) being the most prevalent injury followed by iliotibial band syndrome (ITBS) (3). Furthermore, both PFP and ITBS, along with tibial stress fractures, occur with greater frequency in females than males (3). In a retrospective study of 2002 running injuries, 62% of all subjects diagnosed with PFPS or ITBS were female and 57% of all subjects diagnosed with a TSF were female (3). Further evidence for this sex discrepancy was highlighted in a 2.5 year prospective examination of 1525 Naval recruits where it was observed that females were 2.23 times more likely to develop PFPS than males (4). Several explanations have been offered in the literature to help interpret the sex discrepancy in overuse injury rates, one of which includes the biomechanical differences during gait (5).

### Running Kinematics

It has been suggested that abnormal kinematics during gait may play a role in the aetiology of overuse injuries (6). In light of the sex discrepancy observed in overuse injuries and the idea that abnormal kinematics may be a factor, Ferber et al. found that healthy females display greater peak hip adduction during the stance phase of running when compared to their male counterparts (7). Considering that females diagnosed with PFP, ITBS, and TSF also demonstrate excessive hip adduction during running (8-10) perhaps the altered hip kinematics observed in healthy female runners offers a partial explanation for their elevated risk of injury.

By definition hip adduction (HADD) is the movement of the femur in the frontal plane towards the midline (11). Normally, at initial contact the hip is in either a neutral or slighted adducted position. From there the hip moves into adduction (as the center of mass lowers) before achieving a peak angle at mid-stance (12). The hip then begins to abduct and returns to a slightly adducted position at the end of stance (12). On average peak hip adduction during running in females is approximately 10-17 degrees (7, 13, 14).

It is also important to consider the motion of the pelvis since hip adduction is measured as the angle of the femur relative to the pelvis. Therefore excessive hip adduction can be caused by a lowering of the contralateral side of the pelvis toward the femur. During normal gait, the pelvis is obliquely aligned with the stance (ipsilateral) side slightly elevated and the contralateral side slightly lower at initial contact (15). The contralateral side then continues to lower before reaching a peak angle of approximately 10° during stance (15). After the peak angle has been reached, the contralateral pelvis begins to rise before returning to an approximately neutral position at toe-off (15).

#### Evidence for Hip Kinematics in the Aetiology of Running Injuries

As mentioned in the previous section, there is evidence to suggest that increased hip adduction is a common denominator amongst subjects with PFP, ITBS, or TSF. On average a 2-4° difference in peak hip adduction seems to separate the injured from the healthy controls. Noehren et al., observed that females with PFP demonstrated 2.2° greater peak HADD compared to healthy females (20.0° vs. 17.8°) (14). Similarly, females with a history of TSF ran with 4° greater peak HADD compared to controls (11.7° vs. 7.7°) (8). Lastly, a prospective study of female runners reported that those who eventually developed ITBS ran with 3.1° greater peak HADD compared to healthy controls (14.1 vs. 10.6) (9). While the first two studies mentioned above were cross-sectional in nature it is important to note that the ITBS study was prospective in nature. A similar finding was also observed prospectively in a PFP study where those who went on to develop the injury displayed a 4° increase in peak hip adduction compared to

healthy controls (10). These prospective studies are powerful in that they indicate excessive HADD was present in the runners prior to their injury, and was not simply an antalgic gait response to pain following the injury. They provide stronger evidence to suggest that altered kinematics may be responsible for these overuse injuries.

#### Theoretical Explanation for the Link between Altered Hip Kinematics and Injury

It is important to note that the lower extremity acts as a kinetic chain where the movement of each bone and joint has the ability to influence the movement and loading of adjacent segments and tissues. Abnormal kinematics have been implicated as a potential injury mechanism, where changes in kinematics may alter how a tissue is loaded thereby contributing to the development of injury (8, 9). For example, due to the anatomical location of the IT band (ITB), it has been suggested that excessive hip adduction could increase the strain on the ITB (9). In support of this theory, work by Hamill et al. has shown that runners who incur ITBS, not only demonstrate increased hip adduction during running, but also exhibit an increased strain rate and greater overall strain on the ITB (16). Excessive hip adduction may also theoretically impact loading at the patellofemoral joint since the ITB also inserts into the lateral side of the patella. Increased tension on the ITB has been shown to cause increased lateral translation and tilt of the patella (17). This, in turn, could alter how the contact stress is distributed on the patella potentially leading to the development of PFP. While the literature provides a theoretical explanation for how altered hip kinematics may lead to the development of overuse injuries, the causes of excessive hip adduction during gait remain unclear.

## Development of Altered Hip Kinematics

It has been suggested that altered kinematics during gait may be the result of muscular weakness (18), where the external moments generated during running are greater than the internal moments generated by the musculature. Therefore, the musculature is unable to control the motion of the joint and its associated bones resulting in altered kinematics. In support of this theory, weakness of the hip abductor muscles has been associated with a Trendelenburg gait, which is characterized by a drop in the contralateral pelvis during the stance phase of walking/running (6). Considering that this motion would result in greater hip adduction, it seems pertinent to investigate the relationship between the strength of the hip abduction musculature and hip/pelvis kinematics during gait.

## Hip Musculature

The hip abduction musculature is a group of muscles composed of the gluteus medius, gluteus minimus, tensor fascia latae, piriformis, sartorius, and the rectus femoris (19). While the latter three muscles are considered secondary hip abductors, the former three are referred to as the primary abductors (20) with the gluteus medius providing the greatest contribution to hip abduction (21). The gluteus medius originates on the wing of the ilium and inserts into the lateral and superior-posterior aspects of the greater trochanter (19). It has been labelled as the primary frontal plane stabilizer for it composes 60% of the total physiological cross-sectional area (PCSA) of all of the muscles which assist in abducting the hip. Also due to the location of the muscle(s) with respect to the hip/pelvis it has the greatest abduction moment arm of all of the muscles (19). It consists of three distinct bands of muscle (anterior, medial, and posterior) which function together to stabilize the pelvis and the femur (22). The anterior band displays almost vertical fiber orientation, the largest moment arm in the transverse plane, and a large physiological cross-sectional area which results in the ability to produce a large abduction torque about the hip (19, 21). The middle band displays vertical fiber

orientation, a large transverse plane moment arm, and a large PCSA allowing it to generate a large abduction torque which facilitates pelvic stability (19, 21). The posterior band displays fibers in parallel with the neck of the femur, a smaller moment arm in the transverse plane, and a smaller PCSA; this orientation facilitates its role as a stabilizer of the femoral head within the acetabulum (21, 22). During gait, the three portions fire separately but they maintain a similar pattern of activity with two distinct bursts (21). The distinct innervation (by the gluteal nerve) allows each band to fire at different points of the gait cycle to optimize function (21, 22). This is evident in the anterior band which displays a delayed second burst compared to the middle and posterior bands (21). This is theorized to aid in the rotation of the contralateral pelvis during mid-late stance (21). Furthermore, the unique orientation of the fibers of each band allow them to be optimized for their different functions (21). This can be observed in the posterior band where the muscle fibers are oriented in parallel with the neck of the femur, so it acts as a hip extender as well as an external rotator (19).

It should also be noted that joint position affects the moment arms of the musculature of the hip, and thus affects the resultant muscle torque (19). Therefore, depending on the orientation of the hip joint, muscle function may be altered. For example, as the hip moves into flexion, the posterior portion of the gluteus medius no longer generates an external rotation moment but instead generates an internal rotation moment (19). More pertinent to the stabilization of the pelvis however, the angle of hip abduction has the ability to influence the torque generation of the hip abductors with the greatest torque being produced at -10 degrees of hip abduction (10 degrees hip adduction) and the least torque being produced at 40 degrees of hip abduction (19). This is an essentially negative linear relationship and it is interesting to note that the greatest torque is generated at a joint position which corresponds with the position of the hip joint during the stance phase of walking (19).



## Hip Strength and Kinematics

The link between hip function and injury is supported by research which has shown that individuals with PFP (23-25) or ITBS (26) also present with reduced strength of the hip abduction musculature. In a cross-sectional study by Ireland et al., it was observed that females with PFP demonstrated a 26% reduction in peak isometric H ABD strength compared to healthy controls (24). This is consistent with other literature as strength deficits range from 8-26% in subjects with PFP (12, 23-25). Similarly in females with ITBS, a 20% reduction in peak isometric H ABD strength has been observed between the injured and the un-injured limb (26). Given the role of the hip abduction musculature in controlling hip adduction, it has been suggested that a reduction in strength of this muscle group may contribute to the excessive hip adduction observed in runners with overuse injuries (6).

## Causes of Hip Abductor Muscle Weakness

Despite documentation associating hip muscle weakness and overuse injuries, a definitive cause for this weakness remains unknown (27). It is possible that reduced H ABD strength may be the result of the pathology (6, 24, 27), and it has been suggested that patients with PFP may alter their mechanics in order to reduce pain (28). Consequently this could result in a change in how the muscle is loaded which could alter its function. However, it also seems feasible that the reduced hip abduction strength may have preceded the injury. In the latter case, it seems critical to discern what factors may have lead to this decrement in strength.

Considering that when we measure hip abduction strength we are actually measuring the torque produced by the musculature, it has been suggested that an excessive femoral neck-shaft angle may be of influence (29). The femoral neck-shaft angle has the potential to alter the torque output of the hip abduction musculature, since alterations to the angle result in subsequent changes to the moment arm of the hip abduction musculature (29). For instance, a computational modelling study

observed that a 20° increase in the neck-shaft angle resulted in a 26% decrease in the abduction moment arm (30). While this relationship has not been measured in-vivo, this theory has been supported by the observation that females with PFP display an increased femoral neck-shaft angle compared to healthy controls (29). Therefore, it seems possible that the reduced torque generating capacity of the hip abductor musculature may potentially be a consequence of hip anatomy. In light of this, more research is needed to understand the natural variability in the femoral neck-shaft angle and how this influences strength and consequently gait kinematics.

### Neck-Shaft Angle

The femoral neck-shaft angle is defined as the angle at the intersection of the femoral neck axis and the longitudinal axis of the femur (31). Previous studies investigating the NSA have shown that the angle changes with development; as a child grows the neck-shaft angle decreases as the body begins to bear more weight, and individuals usually achieve adult values by adolescence (32). Normative values have been determined for the human population [126.4 (5.7°)] but climate, socio-economic status, and body size have all been shown to influence the magnitude of the NSA (32). Compared to individuals living in warm climates (Southern Pacific), individuals from North America display a reduced angle (130° vs. 125°) (32). This is thought to be due to the selection pressure for cold adaptation resulting in an increased body mass. This increase in body mass places a greater load on the femoral neck during development and the femoral neck adapts to this increased load by reducing the neck-shaft angle (32).

Further developmental factors influencing the femoral inclination angle also include subject height and pelvis width (33). Early observations of the characteristics of the femoral neck have shown that the neck-shaft angle is the lowest when the bones of the lower limbs are short (subject is short in stature) and when the pelvis is wide (33). These structural characteristics alter how the femoral neck is loaded which consequently affects the NSA development. Specifically, in 1889, Humphry observed

that when the femurs of his cadavers were less than 18 inches long (average = 16.5in), the average NSA was 122.5°. However, when the femurs were longer than 18 inches (average = 19in) the NSA was 125° on average. As Humphry stated: “this difference is to be expected, because the elevation of the pelvis above the knee, together with the narrowness of the pelvis, opens up, as it were, the angle of the neck of the thigh-bone with the shaft” (33). Finally, Humphry postulated that when the pelvis is wide, the weight of the body falls more upon the inside of the epiphyseal line which represses the growth of the neck thereby resulting in the maintenance of a large neck-shaft angle (33). Not surprisingly, considering these developmental factors a wide range of values have been observed in normal human femora from North America with the smallest being 109° and the largest being 142° (32). However it is generally accepted that normal angles for the femoral neck-shaft are between 125-135° (20).

#### Measuring Neck-Shaft Angle

To measure the femoral neck shaft angle (NSA) previous studies have utilized dual-energy X-ray absorptiometry (DXA) (34-36). This technique utilizes the principles of X-ray spectrophotometry and employs beams of two different energy levels to produce images of high resolution (37). The advantages of this system include short scan times, low radiation dose, and rapid patient set-up (37, 38). To measure the NSA subjects are placed supine on the scanning bed and an anteroposterior (AP) radiograph of the hip is taken (35). However, considering that the femoral neck shaft angle (NSA) is a planar angle, rotation of the femur along its longitudinal (vertical) axis has the ability to influence the measurement of the NSA. Therefore, it is important to consider the position of the femur when setting up a scan. Previous research in this area has determined that the optimal position to measure the NSA is with the femur in an internally rotated position (35). Kay et al. (35) compared the measured neck shaft angle of an adult cadaver as it was placed in varying degrees of internal and external rotation, and used this data to construct a mathematical model to predict the NSA of femurs with varying structural characteristics (femoral neck anteversion and neck shaft angle). Their

results indicate that in order to accurately measure the femoral NSA the femur should be placed in an internally rotated position between 10° and 20° IR. External rotation of the femur should be avoided, for as little as 7° ER can result in greater than 10° change in measured NSA (35). Therefore, this demonstrates the importance of standardizing the patient position in order to accurately and reliably measure the femoral neck-shaft angle.

#### Other Structural Measures That May Influence Hip and Pelvis Kinematics

As touched upon earlier in discussion of the femoral neck shaft angle, the width of the pelvis with respect to the length of the femur is speculated to potentially alter the alignment of the lower extremity. Specifically, Horton and Hall postulated that an increased hip width in conjunction with a relatively short femur would place the femur in a more obliquely oriented position relative to vertical, thereby placing the hip joint in a more adducted position (39). Separate studies have reported that females demonstrate an increased pelvis width-femur length ratio (7, 40) and greater hip adduction during running compared to males (7). To our knowledge only one study has measured both the pelvis width-femur length ratio and hip kinematics during running (41). Willson et al. found no difference in pelvis width-femur length between females with and without PFP despite observing a 3.5° increase in hip adduction across a range of activities in the PFP group (41). However, they did not specifically explore whether the pelvis width-femur length ratio has any influence on lower extremity biomechanics. Therefore, further investigation is required to clarify this relationship.

#### Summary

In conclusion, it seems important to study the influence of hip structure on hip strength and consequently kinematics. Given the inherent variability that exists within hip structure and its theoretical ability to alter both muscle moment arms and lower extremity alignment, understanding these relationships may help explain the

mechanisms causing abnormal hip and pelvis motion during running. The findings of such a study would have implications for a number of lower extremity overuse injuries.

## Chapter Two: Introduction

Within the literature it has been well documented that females are more prone to knee injuries than males (42-44). While a conclusive mechanism has not been determined for this discrepancy it has been suggested that abnormal gait kinematics may affect one's propensity for injury (18). In support of this theory, research has shown that females exhibit greater peak hip adduction while running (7). Furthermore, increased hip adduction during running has been associated with several lower extremity overuse injuries. For example, excessive hip adduction has been reported in female runners demonstrating patellofemoral pain (PFP), tibial stress fracture (TSF), as well as iliotibial band syndrome (ITBS) when compared with healthy controls (8-10). The link between hip function and injury is further supported by research which has shown that females with PFP present with reduced isometric strength of the hip abductor musculature when compared to healthy females (8, 9). Given the role of the hip abductor muscles in controlling hip adduction during gait, it has been suggested that a reduction in strength of this muscle group may contribute to the excessive hip adduction angles seen in runners with overuse injuries (6).

Despite documentation associating hip muscle weakness and overuse injuries, a definitive cause for this weakness remains unclear. While hip weakness may be a result of PFP, it is also possible that reduced hip strength may precede injury (6, 24). If the latter is true, it is of particular to discern what factors might lead to this decrement in strength. Specifically, it has been proposed that abnormal hip structure may be of influence (29). It has been suggested that an excessive femoral neck-shaft angle (NSA) influences muscle strength by altering the muscle moment arm (29). This relationship between femoral neck-shaft angle and hip abductor muscle moment arm was explored using a computational modeling approach (30). The aforementioned study demonstrated that a 20° increase in femoral NSA resulted in a 26% reduction in the gluteus medius moment arm (30). This shortening of the gluteus medius moment arm would result in a reduction of the torque generating capacity of the muscle which could potentially make it more difficult to control motions of the lower extremity. Overall, this

may induce abnormal gait kinematics which, as previously stated, has been suggested as a mechanism for PFP (18). This theory has been supported by previous research that found that females with PFP presented with an increased femoral NSA when compared to healthy controls (29). Since the femoral NSA will vary amongst females (32), this may result in observable differences in pelvis/hip frontal plane kinematics, possibly due to the reduced capacity of the hip musculature to control motions of the pelvis/hip. However, no study has specifically explored the relationship between femoral NSA, hip muscle strength, and consequently frontal plane hip/pelvis kinematics. This also necessitates investigating the relationship between hip strength and hip/pelvis kinematics, for if we expect the femoral NSA to influence hip/pelvis kinematics via its influence on hip strength, strength must be related to kinematics.

It has also been suggested that the structural anatomy of the hip and pelvis has the potential to influence lower extremity kinematics (and patellofemoral mechanics) by altering alignment of the lower extremity (18, 29). Specifically, it is the ratio of the width of the pelvis relative to the length of the femur [pelvis width – femur length ratio (PW-FL)] that has been implicated. Previous research, by Horton and Hall, found that females have a larger PW-FL ratio than males (39). This would result in increased hip adduction as the femur must be placed in a more oblique position in order to maintain a normal stance width (39). This increase in hip adduction, as previously stated, has been documented in females exhibiting PFP as well as other lower extremity pathologies (7-10). Despite the potential relationship between hip structure and frontal plane kinematics, at this time, no study has investigated how alterations in structure affect frontal plane kinematics.

In summary, it appears important to evaluate the relationship between structural measurements, muscle force output, and frontal plane kinematics. Studying these relationships may help to further elucidate the theoretical mechanisms associated with the aetiology of certain overuse injuries. Therefore, the purpose of this study was to investigate the relationships between hip structure, hip abduction strength, and frontal plane hip kinematics during running in healthy active females. It was

hypothesized that: (i) a greater femoral inclination angle would be associated with both greater hip abduction strength and hip adduction during running; (ii) a greater pelvis width-femur length ratio would be associated with greater hip adduction during running.



## Chapter Three: Methodology

### Experimental Design

This study utilized a correlational single group design. It was descriptive in nature and used analog observation in order to capture subject behavior. The independent variables included femoral inclination angle, pelvis-width - femur length ratio, and hip abduction strength. The dependent variables included: hip abduction strength, and the following discrete kinematic variables: peak hip adduction, hip adduction excursion, peak contralateral pelvic-drop, and contralateral pelvic-drop excursion. All variables were measured once per subject.

### Subjects

#### *Subject Inclusion/Exclusion*

Using the method described by Watkins et al., an a priori sample size of twenty-two was calculated using a power level of 0.8 and an anticipated  $r$  value of 0.5 (45). Twenty-five female subjects were recruited using convenience sampling, from the university campus and surrounding area through physical and web-based flyers as well as word of mouth. Subject demographics are presented in Table 3.1. below. In order to be included in the study subjects had to be regularly engaged in recreational or competitive physical activity involving running for at least 30 minutes, three times per week. Subjects were excluded from participating if they were: pregnant, not comfortable running on a treadmill without the use of handrails, currently experiencing pain during running, had suffered an injury to the lower extremity and/or back/spine that limited activity in the past three months, or had undergone surgery to the lower extremity/back/spine for any past injury.

Table 3.1. Subject Demographics

	<b>Mean</b>	<b>SD</b>
Age	29	6
Height (m)	1.67	0.05
Weight (kg)	61.6	11.3

### Procedures

Subjects were required to visit the university campus on one or two occasions for approximately 60 minutes on the first visit and approximately 30 minutes for the second. The first visit comprised of strength testing and a gait analysis, and the second visit consisted of pregnancy screening, anthropometric measurements, and a DXA scan. Prior to participation, all procedures were explained to the subjects and they were asked to provide informed consent using a form approved by the Institutional Review Board of the University of Kentucky.

#### *1<sup>st</sup> visit - Biodynamics Laboratory*

During their time in the Biodynamic Laboratory subject data was collected regarding anthropometrics, muscle strength, and running gait. At the beginning of the data collection subjects were asked to change into running shorts and a tank top and all subjects were given a pair of neutral running shoes (New Balance, R662WSB, Boston, MA, USA) to wear during the data collection. Each subject's height and weight was then measured using a balance and stadiometer.

## Strength Testing (Muscle Force Measurement)

The strength testing was performed on an isokinetic dynamometer (Biodex Medical Systems, Shirley, NY) with the subject placed in a side-lying position on a plinth, as described by Jacobs et al (46). The subject's testing limb was placed in a neutral position ( $0^\circ$  of flexion,  $0^\circ$  of abduction,  $0^\circ$  of rotation), the axis of the dynamometer was aligned with the hip joint center in the frontal plane, and the lever arm of the dynamometer was set so that the top of the resistance pad was located 5cm proximal to the knee joint line. Subjects were allowed 2 sets of 3 practice trials to become accustomed to the testing position after which anthropometrics were measured to provide subjects with a period of rest. Subjects were then asked to perform 3 maximal isometric voluntary contractions of their hip abductors (side-lying leg raise) lasting 5 seconds each with 10 seconds of rest in between. Subjects were instructed to slowly ramp up their force production, so that they were applying maximal force by the 3<sup>rd</sup> second of the trial. Verbal encouragement was provided throughout strength testing.

## Anthropometric Measurements

Height and weight were measured for all subjects using a balance and a stadiometer. Pelvis width and femur length were measured with the patient in a supine position on a plinth. Pelvis width was measured as the inter-ASIS distance, and femur length was measured as the distance from the most prominent aspect of the greater trochanter to the knee joint line.

## Gait Analysis – Subject Preparation

Retro-reflective spherical markers (10mm diameter) were placed bilaterally on the lower extremity and the pelvis. Anatomical markers were placed bilaterally on bony landmarks on participants in order to define the joint coordinate systems, these

landmarks included: anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), medial and lateral knee, medial and lateral malleoli, first metatarsal head, and the fifth metatarsal head. Tracking markers were placed on the participants in order to track the movement of segments through space. Tracking markers were placed bilaterally in the following locations: ASIS, PSIS, L5-S1, lateral heel, proximal heel, distal heel, and the most distal point of the toe box. Additionally, one marker was added to the right shoe as an offset to differentiate sides, and clusters of three or four non-collinear markers were placed on the posterior/lateral aspect of the distal thigh and shank (Figure 3.1).



Figure 3.1. Anatomical Marker Set

## Gait Analysis

The gait analysis was completed using 10 Eagle Motion Analysis cameras (Motion Analysis Corp, Santa Rosa, CA) recording at 200Hz and an instrumented treadmill (Bertec, Columbus, OH) recording at 1000Hz. Initially, a static image was captured to identify the locations of the anatomical markers; after which, some of the anatomical markers were removed so that only the tracking markers remained. Subjects were asked to run at a standardized speed (2.7 m/s) and were provided with 3 minutes to warm up and become accustomed to the treadmill condition and speed prior to collecting data (47). Upon completion of the acclimation period, subjects were asked to maintain the same speed while marker trajectory data was collected for two trials of twelve seconds in duration. The total duration of the gait analysis was approximately five to six minutes.

## *2<sup>nd</sup> Visit – Functional Assessment and Body Composition Core Laboratory*

### Pregnancy Testing

In accordance with University of Kentucky procedures and policy, all women of reproductive status underwent a urine pregnancy test immediately prior to dual energy X-ray absorptiometry (DXA) scanning. Only women with a negative urine pregnancy test (within established urine specific gravity ranges) were permitted to undergo the DXA scanning.

### DXA Derived Structural Measurements

Each subject underwent a dual proximal femur DXA scan to provide the skeletal measures for subsequent structural analyses (femoral neck-shaft angle). The anthropometric and DXA scans measures were performed in the University of Kentucky

Functional Assessment and Body Composition Core Laboratory (FAABC) of the Clinical Services Core (CSC) of the Center for Clinical and Translational Science (CCTS) inpatient suite (5th floor Chandler Medical Center).

The dual proximal femur DXA scans was performed using a Lunar iDXA (Lunar Inc., Madison, WI) bone densitometer. The subjects were instructed to remove all objects such as jewelry or eyeglasses and only wore a standard hospital gown or t-shirt and shorts containing no metal during the scanning procedure. In order to ensure accurate scans were taken of the pelvis, all scans were taken from an anteroposterior view with the subject's legs placed in 20° of internal rotation (35). Internal rotation was standardized for all subjects using a triangular block (with a fixed angle of 20°) positioned at the subjects feet. Once the block was in position, the subject's leg was moved into position by simultaneously rotating the foot, tibia, and femur. Finally the subject's foot was fixed to the block using a Velcro strap and the subjects were instructed to maintain the position while the scan was performed. All scans were analyzed by a single trained investigator using the Lunar iDXA enCORE software version 14.10.022 (Lunar Inc., Madison, WI)

#### Data Analysis

All marker trajectory data were collected and tracked using Cortex software (Motion Analysis Corp, Santa Rosa, CA) while all processing was performed using Visual 3D. All kinematic data was filtered using a fourth order low-pass Butterworth filter with a cut-off frequency of 8Hz. A cut-off frequency of 8Hz was used in order to be able to compare our results with previous literature (8, 14, 48). An x-y-z (medio-lateral, antero-posterior, vertical) cardan series of rotations was applied to quantify joint angles where the distal segment was referenced to the proximal segment. Foot strike was identified

as the point at which the vertical velocity of the distal heel marker changed from positive to negative and toe-off was identified as the point of peak knee extension (49). This method was chosen due to errors in force data recorded for some subjects. Discrete variables of interest (peak hip adduction, hip adduction excursion, peak contralateral pelvic-drop, contralateral pelvic-drop excursion) were extracted from the processed data using Visual 3D (C-Motion Inc, Germantown, MD. The kinematic variables of interest were defined as follows: peak angle was the greatest value observed during stance, excursion was the difference between the peak angle and the angle observed at heel-strike. With respect to contralateral pelvic drop, a negative joint angle indicated contralateral pelvic drop while a positive value indicated contralateral pelvic rise.

Structural measurements of the pelvis/femur, derived from the DXA scans, were measured using Lunar iDXA enCORE software 14.10.022. The femoral neck-shaft angle was defined as the angle between the axis of the neck of the femur (length from below the lateral aspect of greater trochanter to the caput femoris) and the longitudinal shaft of the femur (29, 34) (Figure 3.2). The pelvis-width to femur length ratio was defined as the quotient of the pelvis width (measured as the distance between left and right anterior superior iliac spine (ASIS) and the femur length (measured as the distance from the greater trochanter to the ipsilateral knee joint line) (40).

Strength measurements were quantified using the torque output from the Biodex. The outputted torque was converted to force by dividing out the length of the lever arm of the Biodex, and the force values were then multiplied by the subject's femur length and normalized to each subject's body weight. Additionally, in order to compare results to the PFP literature the force produced by each subject (calculated as previously mentioned) was normalized to each subject's body weight in newtons (23, 24, 50).

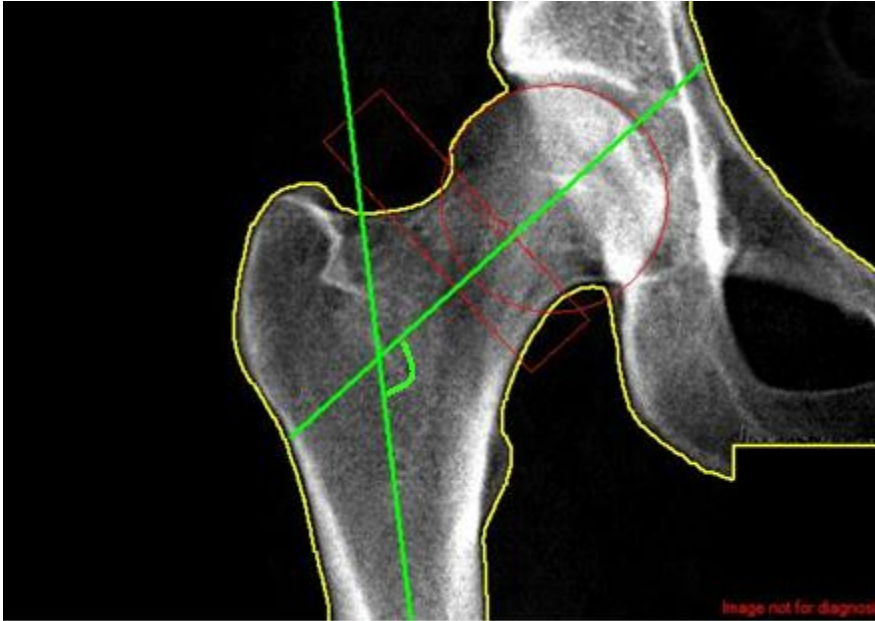


Figure 3.2. Femoral Neck-Shaft Angle

#### Statistical Analysis

The association between the strength measurements, the kinematic variables of interest, and the structural measurements were assessed using Pearson correlations. Correlations were performed for femoral NSA vs. HADD strength, femoral NSA vs. HADD and contralateral pelvic drop, HADD strength vs. HADD, and PW-FL vs. HADD and contralateral pelvic drop. Interpretation of Pearson correlation coefficient,  $r$ , followed the guidelines set out by Watkins et al. where a good to excellent relationship was defined as  $r > 0.75$ , a moderate to good relationship as  $r = 0.5 - 0.75$ , a fair degree of relationship as  $r = 0.25 - 0.50$ , and little or no relationship as  $r = 0.00 - 0.25$  (45). Statistical Significance was defined as  $P < 0.05$ .

To assess the intra-rater reliability of the measurement of the femoral NSA, 10 scans were analyzed on two occasions by a single investigator to determine the intraclass correlation coefficient (1, 1) and the standard error of the mean (SEM). Additionally, 5 scans were analyzed by two investigators in order to measure the inter-



rater reliability using the intraclass correlation coefficient (3, 1) and the SEM (51). All statistical analyses were performed using SPSS statistical software (SPSS Inc, Chicago, Illinois).

## Chapter Four: Results

Subject's mean values for femoral NSA, PW-FL, and isometric H ABD strength are presented in Table 4.1. A reliability analysis performed on the femoral NSA data demonstrated that the angle could be measured reliably with an intra-rater correlation coefficient [ICC (1, 1)] of .970 and a SEM of 1.17° and an inter-rater correlation [ICC (3, 1)] coefficient of .968, and a SEM of 1.22°.

**Table 4.1.** Femoral Neck-Shaft Angle (NSA), Pelvis Width-Femur Length Ratio (PW-FL), and Hip Abduction (ABD) Strength

	<b>Mean</b>	<b>SD</b>
<b>Structure</b>		
NSA (°)	131.0	6.8
PW-FL Ratio	0.545	0.048
<b>Isometric Strength</b>		
Hip ABD (Nm/Kg)	0.86	0.18
Hip ABD - %BW (N/N)	24	5

Furthermore, a fair degree of relation was observed between femoral neck-shaft angle and side-lying isometric hip abduction strength (Table 4.2; Figure 4.1) where a larger femoral neck-shaft angle was associated with lower peak isometric hip abduction torque.

Table 4.2. Relationship between Femoral NSA, Max Hip Abduction Strength (ABD), Peak Hip Adduction (ADD), Hip Adduction Excursion (EXC), Peak Contralateral Pelvic Drop, and Contralateral Pelvic Drop Excursion

	<b>Pearson <i>r</i></b>	<b><i>P</i> Value</b>
<b>Strength</b>		
Hip ABD	-0.472	0.017 <sup>a</sup>
<b>Kinematics</b>		
Peak Hip ADD	0.331	0.106
Hip ADD EXC	0.189	0.366
Peak Contralat. Pelvic Drop	-0.008	0.969
Contralat. Pelvic Drop EXC	0.02	0.924

<sup>a</sup> Denotes significance at  $P < 0.05$  level

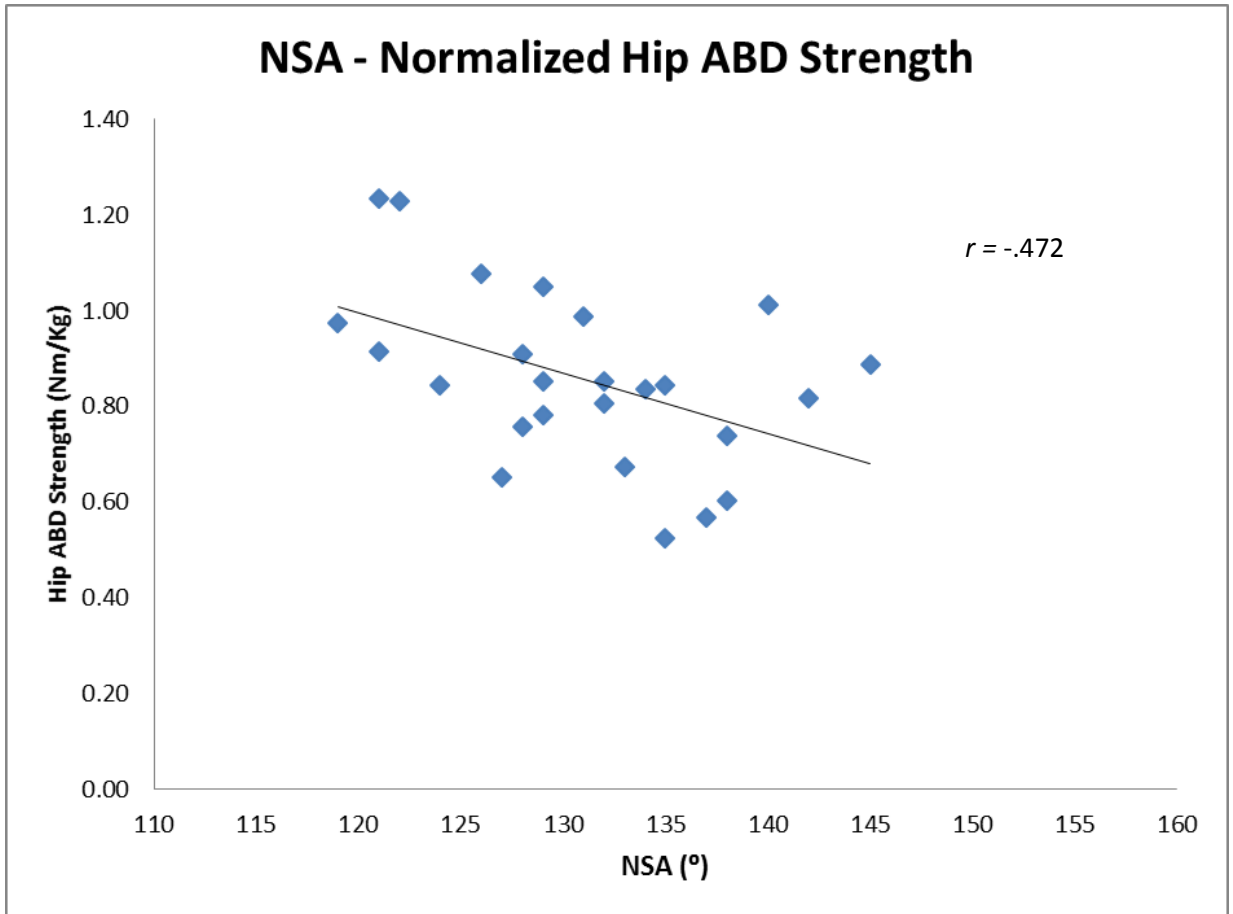


Figure 4.1. Relationship between Femoral NSA and Normalized HABD Strength

During the treadmill running task, on average, subjects underwent 7.9° (3.9) of hip adduction excursion before achieving a peak hip adduction of 17.4° (4.6) (Figure 4.2); while at the same time, subjects underwent 4.1° (2.1) of contralateral pelvic drop excursion before achieving a peak contralateral pelvic drop angle of 5.9° (2.8) (Figure 4.3)

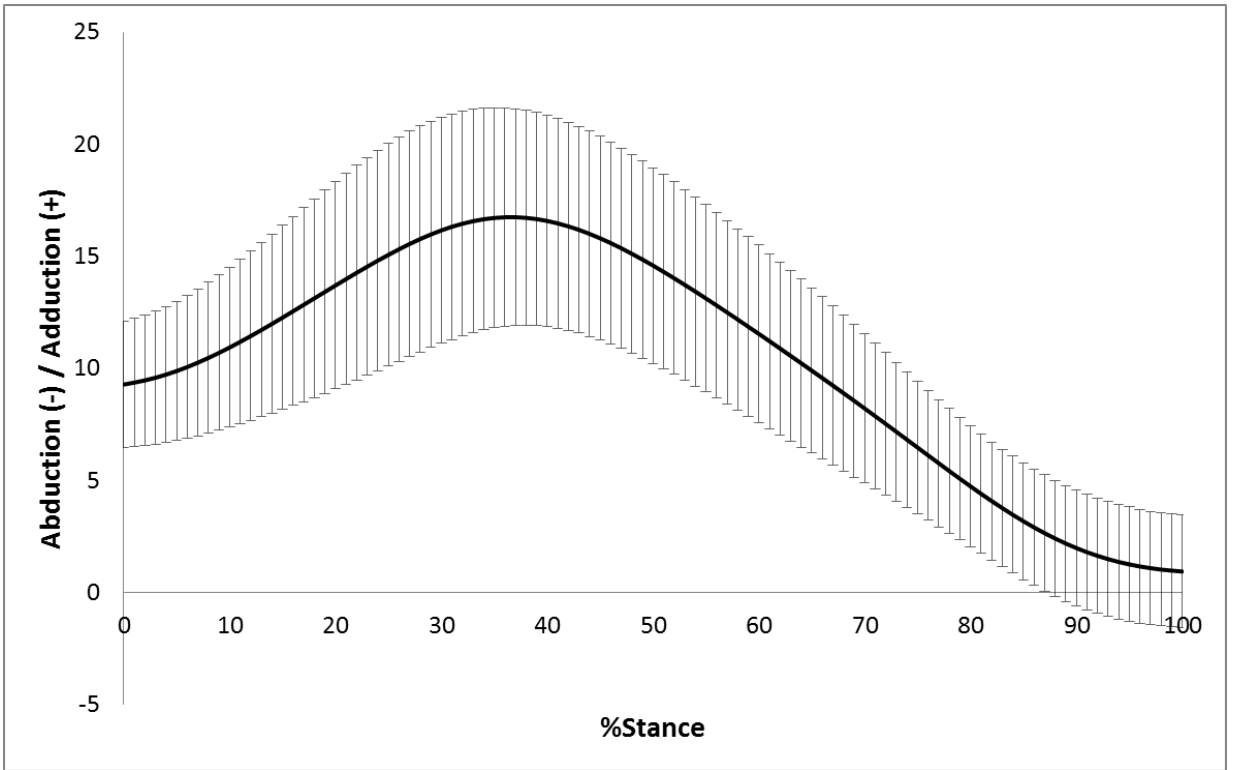


Figure 4.2 Hip Adduction Ensemble Curve (Error Bars Represent  $\pm 1$  SD)

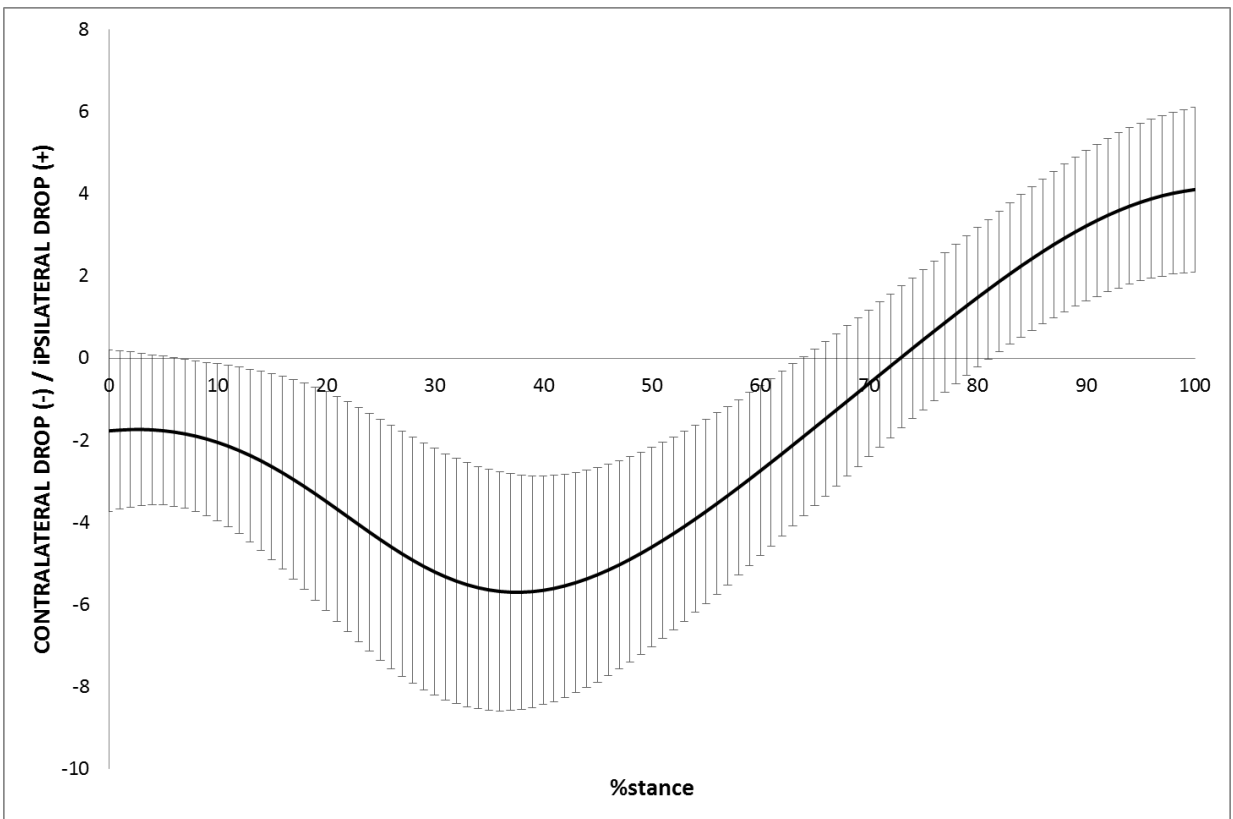


Figure 4.3. Contralateral Pelvic Drop Ensemble Curve (Error Bars Represent  $\pm 1$  SD)

No relationship was observed between femoral NSA and peak hip adduction, hip adduction excursion, peak contralateral pelvic drop, or contralateral pelvic drop excursion (any of the lower extremity frontal plane kinematic variables measured during running) (Table 4.2). Additionally, no relationship was observed between pelvis width-femur length ratio and peak hip adduction, hip adduction excursion, peak contralateral pelvic drop, or contralateral pelvic drop excursion (any of the kinematic variables measured) (Table 4.3).

Finally, no relationship was observed between peak hip abduction strength and peak hip adduction (Figure 4.4), hip adduction excursion, peak contralateral pelvic drop, or contralateral pelvic drop excursion (Table 4.4).

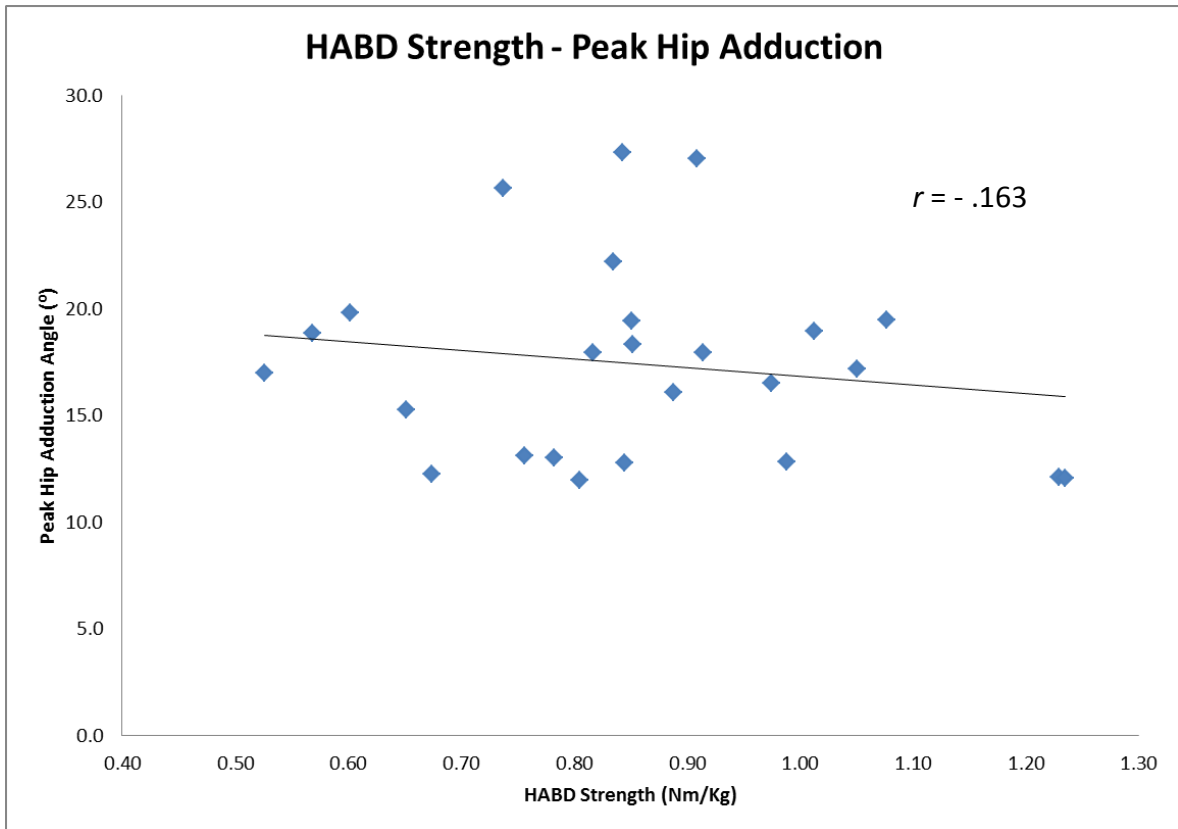


Figure 4.4. Relationship between HABD strength and Peak Hip Adduction During Running

Table 4.3. Relationship of Pelvis Width – Femur Length Ratio with HABD Strength and Kinematic Variables

	<b>Pearson <i>r</i></b>	<b><i>P</i> Value</b>
<b>Kinematics (°)</b>		
Peak Hip ADD	0.054	0.799
Hip ADD EXC	-0.194	0.352
Peak Contralat. Pelvic Drop	0.059	0.778
Contralat. Pelvic Drop EXC	0.084	0.690

Table 4.4. Relationship between Strength and Kinematic Variables of Interest.

	<b>Pearson <i>r</i></b>	<b><i>P</i> Value</b>
<b>Kinematics (°)</b>		
Peak Hip ADD	-0.163	0.435
Hip ADD EXC	-0.119	0.571
Peak Contralat. Pelvic Drop	0.16	0.939
Contralat. Pelvic Drop EXC	0.062	0.769

## Chapter Four: Discussion and Conclusions

### Discussion

The aim of this study was to investigate the relationship between selected hip structure measures, hip abductor muscle strength, and frontal plane hip and pelvis kinematics during treadmill running. Consistent with our hypothesis, a fair relationship ( $r = -.472$ ) was observed between femoral NSA and side-lying isometric hip abduction strength, where an increased femoral NSA was associated with reduced HABD strength. However, inconsistent with the remainder of our primary hypothesis, no relationship was observed between femoral NSA and any of the hip and pelvis kinematic variables measured during running. Furthermore, no relationship was observed between pelvis width-femur length ratio and HABD strength or any of the kinematic variables.

To our knowledge, the only previous study investigating the relationship between the femoral NSA and hip abduction muscle performance was by Arnold et al., who utilized a computer simulated model to investigate the effects of femoral NSA on the gluteus medius muscle moment arm (30). Their findings demonstrated that when the femoral NSA was increased by  $20^\circ$ , the gluteus medius moment arm was reduced by 26%. This would, in turn, lead to a theoretical reduction in the torque generating capacity of the musculature which would need to be compensated for via increased muscle activation. However, their study did not investigate the effects of smaller changes in the femoral NSA. Thus, the results of the present study confirmed in-vivo the results of the modeling study while also demonstrating that 22% ( $R^2 = .221$ ) of the variance in HABD strength can be accounted for by changes in the femoral NSA.

On the other hand, we did not find a relationship between femoral NSA and frontal plane hip/pelvis kinematics, nor did we observe a relationship between HABD strength and frontal plane hip/pelvis kinematics. However, perhaps it is not surprising that the femoral NSA does not influence frontal plane kinematics considering that it only had a small influence on HABD strength. This questionable relationship between strength and kinematics is highlighted in work by Heinert et al., who measured the peak



isometric strength of 150 female subjects and then compared the running kinematics of the strongest and the weakest quartiles. No difference in HADD was observed at either initial contact, peak HADD, or toe-off (52). Further support for the theory that HADD strength may not be a strong predictor of hip kinematics during gait is provided by several studies that have explored the effects of a hip strengthening program on gait kinematics (50, 53, 54). In similar studies by Snyder et al. and Willy et al. no changes in HADD were observed during running despite successfully increasing the HADD strength of healthy individuals (50, 54). This was also demonstrated in subjects with PFP whereby no change in HADD range of motion was observed during running despite increasing HADD strength (53). Therefore, it seems that our observation supports the consensus of the literature that there is little to no relationship between peak isometric HADD strength and HADD during running.

One possible explanation for the lack of relation between strength and kinematics is that running may not be a demanding enough task to highlight strength discrepancies between subjects. Perhaps the hip abductor musculature does not need to fire maximally in order to maintain a normal gait pattern. Therefore, strength discrepancies observed between subjects would not lead to kinematic differences because the weaker subjects still have enough strength to maintain normal gait. This has been observed in walking, where a 24% reduction in strength of the gluteus medius via a gluteal nerve blocking injection resulted in no changes to frontal plane kinematics (55). Considering that many muscles compose the hip abduction musculature perhaps a reduction in strength of the gluteus medius can be compensated for by the remaining muscles. Further research is needed to understand how much force is required of the hip abductor musculature to maintain normal gait patterns.

Another possible explanation for the lack of relationship observed between strength and kinematics is that perhaps measuring peak isometric hip abduction strength is not be the most relevant method to quantify function of the hip abduction musculature during a dynamic weight bearing task such as running (27). Considering that running requires the hip abductors to repetitively work both eccentrically and

concentrically, it may be more relevant to measure the endurance capacity of the musculature rather than the peak isometric force. For example, Souza et al., reported that isotonic hip extension endurance was the only significant predictor of hip internal rotation during running (29). However, further research is required to determine whether hip abductor endurance might be a better predictor of hip adduction during gait compared to the traditional measurement of peak isometric strength.

The findings of the present study may also have clinical implications. It has been proposed that the excessive HADD seen in some PFP patients during running may be the result of weakness of the hip ABD musculature (6, 7). However, our results demonstrate little to no relationship between HADD strength and HADD. Despite evidence from cross-sectional studies demonstrating decreased abduction strength in subjects with PFP, prospective studies have not found a similar relationship and have been unable to demonstrate reduced HADD strength as a risk factor for development (27). In support of this, a recent meta-analysis observed that limited evidence exists to support the idea that reduced isometric hip abduction strength is one of the causes of PFP (27). Therefore, while it is possible that strength deficits may play a role in the aetiology of PFP, they may not be the primary contributing factor to the excessive hip adduction seen in this population.

An additional aim of our study was to investigate the relationship between pelvis width-femur length ratio and frontal plane hip kinematics. Prior to our investigation it was hypothesized that a positive relationship would be observed between pw-fl and hip adduction. Specifically we believed that as the pelvis increased in width a concomitant increase in hip adduction angle would also be observed. However, our results revealed no relationship between pw-fl and hip adduction during running, which suggests that the width of the pelvis (when normalized to leg length) has little to no influence on the magnitude of peak HADD or HADD excursion during running. Previous studies investigating the relationship between anthropometrics and kinematics support our findings. For example, Willson et al., found no difference in pw-fl between females with and without PFP despite observing increased HADD in the PFP group during running

(41). However, in the above mentioned study, all joint angles were normalized to the static pose therefore any influence that abnormal structural alignment would have had on the resultant calculation of HADD would have been negated. However, in our analysis joint angles were not normalized and we still found no relationship between pw-fl and HADD. Considering these findings, perhaps the body is able to adapt to these structural differences and employ altered neuromuscular control in order to compensate. The benefits of altered neuromuscular control can be observed in studies which have successfully reduced the magnitude of HADD during running by employing neuromuscular re-education via. gait re-training (56, 57). Therefore, perhaps a relationship was not observed between pw-fl and HADD due to adaptations specifically employed to compensate for structural differences.

#### Limitations

Several limitations should be noted in this study. First, while all marker placement was performed by a single trained investigator it should be noted that the possibility of marker placement error is always present. Further errors may be present due to skin movement artefact where the markers lying on top of the skin do not represent the underlying bony movement. This is an unavoidable limitation of 3-D motion analysis; however considering the subject pool consisted of physically active individuals the magnitude of this error would be smaller than if our subjects were primarily overweight or obese. In addition to errors present in motion analysis, it is possible that errors may be present in strength testing data due to submaximal exertion by the participants. Furthermore, despite standardizing the patient position for the DXA scan, the possibility exists that all subjects were not in the same position due to differences in structural alignment which could induce errors into the resultant calculation of the femoral NSA. Further improvements in scanning technology where the plane of the femoral NSA could be aligned with the imaging plane would allow for more accurate measurement of this angle. Finally, considering the small sample size and

that all subjects were healthy, caution should be taken when attempting to infer these results across different patient populations. Future research could build on our work by assessing the relationship between femoral NSA, hip strength, and hip kinematics in an injured population.

## Conclusion

In conclusion, the results of the present study demonstrate that a fair relationship exists between the femoral NSA and peak isometric HADD strength; where, as the angle increases in magnitude a concomitant decrease in peak isometric HADD strength is observed. However, no relationship was observed between the femoral NSA and peak HADD, and no relationship was observed between HADD strength and peak HADD. Lastly, no relationship was observed between pw-fl and peak isometric HADD strength or peak HADD.

Our findings suggest that the magnitude of hip adduction observed in healthy females during running is not the result of weak hip abductors or deviations of the neck shaft angle. This questions whether the excessive hip adduction seen in female runners with overuse injuries is caused by decrements in strength and hip structure. Further research is needed to determine whether there are other underlying factors causing excessive hip adduction.

## Appendix A: Model Definition Template

### **Foot:**

The superior-inferior ( $z$ ) axis was defined as the vector from the distal heel to the proximal heel. The antero-posterior axis ( $y$ ) was parallel to the floor and was defined as the vector from the distal heel to the midpoint of the first and fifth metatarsal heads. The cross-product of the two former axes gave the medio-lateral axis ( $x$ ) with the positive direction to the right

### **Shank:**

The frontal plane was defined using the medial and lateral knee markers and the medial and lateral malleoli. The vertical axis ( $z$ ) ran in the direction between the midpoint of the malleoli and the midpoint of the femoral condyles with positive defined as proximal. The antero-posterior axis ( $y$ ) was perpendicular to the plane formed by the femoral condyle and malleoli markers with the anterior direction positive. The cross-product of the two former axes gave the medio-lateral axis ( $x$ ) with its positive direction to the right.

### **Thigh:**

The frontal plane was defined using the hip joint center and the medial and lateral knee markers. The vertical axis ( $z$ ) was defined as the vector starting between the midpoint of the femoral condyles to the hip joint center with its positive direction defined as proximal. The antero-posterior axis ( $y$ ) was perpendicular to the plane formed by the femoral condyles and the hip joint center with its positive direction anterior. Lastly, the cross product of the first two axes defined the medio-lateral axis ( $x$ ) with its positive direction to the right.

**Pelvis:**

A CODA pelvis was used to define the pelvic coordinate system and the hip joint centers (58, 59). The transverse plane (x-y) is defined as the plane passing through the left and right ASIS and the midpoint of the left and right PSIS markers. The medio-lateral (x) axis is defined from the origin (midpoint of ASIS) towards the right ASIS. The vertical (z) axis is perpendicular to the transverse plane. The antero-posterior (y) axis is defined as the cross-product of the y-axis and z-axis.

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