The Effect of Altered Running Form on Overuse Injury Risk Factors

Undergraduate Honors Thesis

Presented in Partial Fulfillment of the Requirements for the Graduation with Distinction in the School of Allied Medical Professions of The Ohio State University

By

Christine M. Glendon

Biomedical Science

The Ohio State University

2010

Thesis Committee:

Christopher McKenzie, Anne Kloos and Ajit Chaudhari, Advisor

Abstract

Running is a popular sport with more competitive participants every year. This rise in popularity has contributed to a rise in prevalence of overuse injuries such as patellofemoral pain, iliotibial band syndrome, and tibial stress fractures. While it is widely believed by coaches and runners that running form, which includes core control and position, plays an important role in injury prevention, little quantitative data exists to support these claims. The impact of altering running form on biomechanical loadings, as well as the best method to achieve changes, is unknown. Previous studies have explored the association of biomechanical loadings with particular overuse injuries, but the impact of core stability and control on those loadings is unknown. Pelvic tilt is a factor in core control and can indicate weakness or abnormal muscle activation in the trunk, and so may also play a role in changing loadings during motion. To explore the association between pelvic tilt and biomechanical loadings associated with running overuse injuries, and to test the effects of one type of technique instruction for running biomechanics, human subjects performed running tasks. Via motion and force capture technology, the gait cycle and loadings of the subjects were collected as they performed normal running and altered pelvic tilt running tasks. Loadings were calculated and normalized by body mass. Subjects were found to have the ability to change pelvic tilt in the anterior or posterior direction. Hip adduction moment and impulse were reduced when subjects altered running form by increasing anterior tilt, and these reductions may correlate to a decreased risk for iliotibial band syndrome. Further study and confirmation of this data is recommended.

Acknowledgments

I would like to thank Dr. Ajit Chaudhari, my advisor, for his support and guidance throughout the project. I would also like to thank Steve Jamison for his assistance in troubleshooting and for writing program codes to aid in data processing and retrieval. Other members of the lab, including Nick Yontz, Natasha Mehta, Ursula Brush, and Josh Cotter, were also essential to this project for their assistance with testing subjects. This study was supported with financial assistance from the NFL Charities Foundation. Finally, I would like to thank Gregory Young for his timely and extremely helpful aid with the statistical analysis of this project.

Table of Contents

List of Tables

List of Figures

Chapter 1

a. Problem Statement

Running overuse injuries are highly prevalent in recreational and serious runners. Running coaches and other experts often rely on anecdotal or experiential evidence for the training and advice given to their athletes, and most coaches provide information about running form. However, efficient running form does not necessarily translate into running form that helps prevent or minimize injury risk. The mechanisms of running injuries are not well understood, and it is widely thought that improved core strength and stability helps to prevent and treat injury, and improve athletic performance. Effective treatment of sports injuries generally requires understanding movement patterns of the athlete and how those movements may lead to injury. Exploring the causes of injuries, why some athletes are more susceptible to particular injuries than others, and methods of identifying biomechanical flaws will improve the scientific knowledge base about injury risk and prevention of running related injuries.

Running overuse injuries frequently occur from abnormal repetitive motion during the gait cycle. Abnormal motion during the gait cycle is often due to muscle weaknesses or imbalances. This may be why physical therapists, athletic trainers, and running coaches commonly believe that strengthening the core muscles leads to more efficient running. Also, better control of the trunk is thought to provide a foundation for controlled movement of the extremities [10]. Little evidence exists to support that claim, so investigation of the effects of core strength and stability on the movement and loadings of the lower extremities is important. A correlation between trunk control and biomechanical loadings that can cause overuse injuries has been suggested [9] but the extent of the correlation, and thus the injury risk, is unknown. Pelvic tilt is a measure of trunk control, as the ability to control the pelvis depends on the muscles of the trunk. Excessive pelvic tilt during motion can indicate weakness or abnormal muscle activation, which may lead to injury [6]. Improved trunk control may allow for faster or more efficient running with less injury risk by influencing the knee and hip loadings during motion. This can be tested by studying the kinematics and kinetics of motion and the biomechanical loads applied to joints during motion. Application of information regarding injury prevention is also an essential part of reducing injury risk. Thus, testing the ability of simple, easily applied methods of prevention such as simple verbal instructions during motion to effect change on a runner's form should be investigated as well.

b. Background

Core Strength and Stability

In recent years, the use of core strength and stability training has dramatically increased in popularity for injury prevention, injury treatment, and improved athletic performance. However, this increased interest in core strength and stability has occurred without strong scientific evidence to support these treatments. Core strength and stability training have been commonly used to prevent and treat low back pain [8], and other uses of similar training are thought to promote injury prevention and improved performance via creating a stronger foundation for motion of the extremities [10]. The terms core strength and stability are often used interchangeably in everyday language, but they are not the same. The “core” of the body is another term for the trunk of the body, and this includes the abdominal, oblique, lower back, and hip musculature. Core strength is the ability to produce or sustain a force for a given amount of time. Core stability is the ability to maintain or resume a position of the trunk in presence of a perturbation [9]. Both are essential for maintaining control of the trunk.

Leetun et. al (2004) found that core strength influences injury risk in athletes, though the study used the term “stability” instead of strength, which is what was really tested in that study. A recent study in the OSU Sports Biomechanics Laboratory suggested that core stability plays a role in biomechanical loads during running [9]. Zazulak et. al (2007) measured core stabilization in collegiate athletes and then tracked injury incidence to show that deficits in core stabilization predict knee injury risk. The relationship between core stabilization and specific biomechanical loadings has not been established, and no evidence exists to show that strength or stabilization of the trunk improves athletic performance. Thus, investigation of the relationship between core strength and stability measures, which includes pelvic tilt, and biomechanical loadings is an underexplored but valuable field of study.

Biomechanical Loadings

Previous research has associated certain biomechanical loadings with increased risk of running overuse injuries. Biomechanical loadings are the forces, moments, and impulses that are applied to specific areas of the body at a given moment during motion. Biomechanical loadings are dependent on body position and can be highly variable. For example, even a small change such as holding a different object during the same motion can change the biomechanical loadings associated with the activity [2]. This is one of the reasons why a change in running form is hypothesized to produce a significant change in the loadings previously associated with overuse injury risk. Those loadings include ground reaction force, peak external hip adduction moment, knee abduction impulses, and internal knee rotation. Overuse injuries are injuries that occur from repeated, abnormal or excessive forces, moments, and impulses over time.

A force is an influence which causes a free body to undergo acceleration. Moments are described as the net result of muscular, ligament, and forces acting to alter the angular rotation of the body [15]. For example, knee adduction moments occur when the tibia is being moved closer to the center of the body. Hip adduction moments occur when leg is being moved closer to the center of the body. An impulse is a moment multiplied by time, so in this case, it is the effort required by the muscles to counteract the externally applied moment multiplied by the time of effort expenditure. Other examples of biomechanical loadings are flexion moments and reaction forces. For example, knee flexion moment occurs when the external world is acting to flex the knee and the body resists. Vertical ground reaction force indicates the magnitude of the vertical force when the foot strikes the ground.

Gait Cycle and Loadings

Loadings discussed here are external loadings, or those that are applied by the environment on the specified anatomical area, unless otherwise specified. The loadings studied are generally at the maximal value during the stance phase of the gait cycle. The gait cycle is the rhythmic, alternating pattern of movement that propels the body forward for walking or running. The gait cycle for running is divided into several phases. A complete gait cycle goes through the stance, float, swing, and float phases on each foot and is generally defined as "the movements and events that occur between two successive heel contacts of the same foot" [15]. The peak biomechanical loadings in the gait cycle include forces, moments, and impulses.

Overuse Injuries and Biomechanical Loadings

The biomechanical loadings essential to this project include knee and hip adduction moments, knee flexion moment, and vertical ground reaction force. These loadings are always at maximal values during the stance phase of the gait cycle. These loadings are of interest as they have been previously associated with the running overuse injuries of iliotibial band syndrome, patellofemoral pain, and stress fractures. Iliotibial band syndrome is characterized by lateral knee pain in runners and is caused by inflammation or tightness in the iliotibial band, a strip of connective fascia that runs from the hip to the knee on the lateral side of the body. Peak hip adduction moments and hip adduction impulse were significantly higher in runners who experienced iliotibial (IT) band syndrome in a previous prospective study by MacMahon et. al [13]. Patellofemoral pain, commonly referred to as "runner's knee", is characterized by pain around or behind the patella and occurs when a mistracking patella irritates the femoral groove and surrounding soft tissue. Patellofemoral pain has been linked with dynamic overloading of the knee and knee flexion moment [16]. Stress fractures are tiny cracks in weight bearing bones, sometimes referred to as "hairline fractures". Previous studies [3, 11] have associated increased ground reaction force with increased risk of stress fractures. Studies such as these have identified dynamic overloading as a primary mechanism for many lower-extremity overuse injuries such as these, but the link between specific biomechanical loads, core strength and stabilization measures and injury risk has not been well-studied.

Technology Utilized

Two special equipment systems were used for this project, a motion capture system and a force plate system. Motion capture allowed researchers to study movement in great detail that is not possible without the technique. Optical motion capture provides information about joint angles, bone movement and positioning, trajectories of movement of body parts. Force plates were used to capture force information and software developed in the lab converts all of this information into meaningful data. Motion capture, combined with the force plate system, also provided researchers an opportunity to study the variability of a subject's movement, to evaluate a subject's balance, or to study many other aspects of a subject's movement.

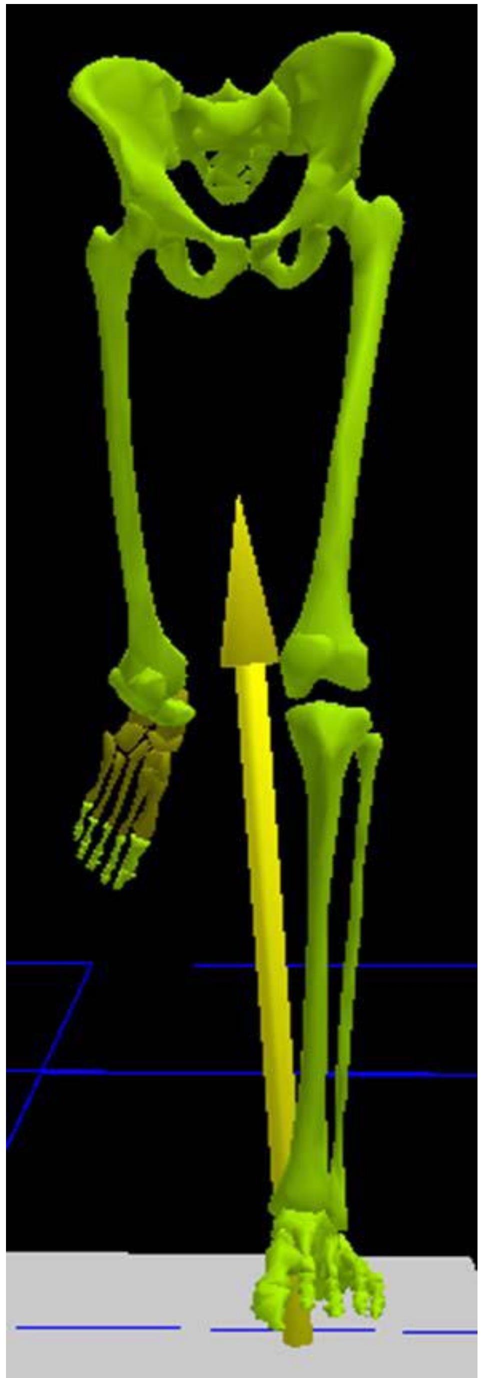

Figure 1. Representation of ground reaction force

indicated by the yellow arrow. Image courtesy of Steve Jamison.

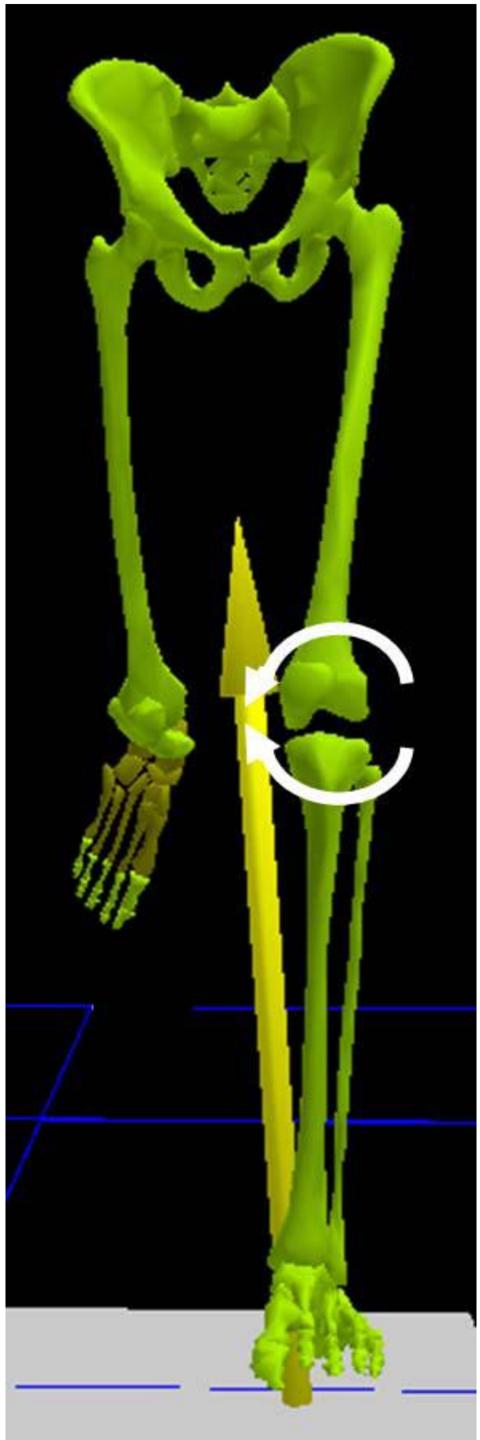

Figure 2. Representation of knee adduction moment

indicated by the circular, white arrows. Image courtesy of Steve Jamison.

c. Objectives

Determine if subjects are able to consciously change pelvic tilt while in motion.

Studying the effect of changes in pelvic tilt on biomechanical loadings in the lower extremity is contingent on subjects' ability to control and change pelvic tilt while running. It is expected that subjects will be able to consciously change pelvic tilt in a significant manner. If subjects are unable to do this, possible explanations include actual inability of subjects to alter pelvic tilt or insufficiency of simple verbal instructions to cause change.

Study the effect of pelvic tilt on knee, hip, and other biomechanical loadings during the stance phase of the gait cycle.

Increased pelvic tilt is expected to be associated with higher loads on the knee and hip, which indicates that increased pelvic tilt correlates with increased injury risk. Pelvic tilt and irregular pelvic motion can indicate previous injury or weak hip abductors [11]. Strength imbalances are associated with increased risk of injuries [12]. Studying the peak loadings on the hip and knee joints in relation to pelvic position can help estimations of relative injury risk at those moments.

Chapter 2

a. Design

This study was designed to investigate the effect of altered running form on the biomechanical loadings previously associated with running overuse injury risk. To achieve this goal, specialized motion capture and force measuring technology was utilized. To study the effect of pelvic tilt on loadings, subjects were asked to run normally and with attempts at anterior and posterior tilt. Subjects served as their own controls to account for variance in individual running style. Three trials of each condition were performed to provide a better data set. Data was collected for dominant foot loadings only to minimize error due to subject differences. Another control for this study was the very specific subject demographics – to participate in this study, subjects had to be males aged 18-24 who were fit, worked out regularly, had no current injuries, and no history of open abdominal surgery. The controlled population helps to reduce the effect of other unforeseen variables but may limit the application of findings to the general population, as the study population does not represent the full range of individuals who might participate in distance running.

b. Methodology

Subjects were recruited from the student population at the Ohio State University and from the general public population surrounding the campus by fliers and word-of-mouth. An Institutional Review Board (IRB) approved flier was circulated in dormitories, recreational facilities, and other campus buildings to recruit subjects.

Prospective subjects were asked to complete an online survey to determine their eligibility for the study. Exclusion factors from the study were current lower extremity injury, previous history of open abdominal surgery, and any other condition that would prevent the prospective subject from being able to run and perform other physical tasks. Subjects had to be able to run comfortably for at least twenty minutes to be considered for the study.

After subject recruitment, subjects came into the laboratory to perform the testing protocol. One subject was tested at a time. The testing protocol included review of documentation, height and weight measurements, the motion test and other tasks specific to the football study. After the initial testing session, subjects' involvement in this study was complete, though involvement in the other study continued. The testing session began when subjects came into the laboratory and filled out forms with contact information. Subjects were asked to review the description of the study and sign appropriate waivers and acknowledgment of risks. Self reported injury histories and foot dominance were recorded. Foot dominance was determined by asking subjects which foot would be used to kick a soccer ball. Height and weight data was collected, and then the subject was asked to change into shorts. Subjects had to be dressed in minimal clothing, as the markers needed for accurate data collection would move too much unless the markers were applied to bare skin.

Subjects were marked with retroreflective markers on anatomical landmarks across the body. A set of 77 markers per subject was used for this study, with most of the markers focused in the lower extremity. The markers were attached to the subject with temporary double sided tape. In the case of excess hair or sweating, which may loosen the tape and cause marker loss, a spray adhesive, Tuf-Skin, was used in addition to the double sided tape. Markers were placed according to anatomical landmarks such as the anterior superior iliac spine and additional markers were placed using a modified version of the "Point Cluster Technique" described by TP Andriacchi in his study on in vivo motion analysis [1].

Figure 3. Marker set, anterior view

Figure 4. Marker set, posterior view

Figure 5. Close-up of PCT marker set

This method helps to account for soft tissue artefact, or the movement of soft tissue and the skin that could otherwise cause errors in the estimation of the motion and loading of the underlying skeleton. The markers allow the Vicon motion capture system and Nexus software to create a three-dimensional representation of the subject's motion by creating a virtual skeleton of the subject.

After marker placement, subjects performed a static calibration trial and photos of subjects were taken for later reference purposes. The subject was then asked to perform dynamic tasks related to the activities of daily living, such as walking and stair climbing, which were not pertinent to this study. Subjects then ran, as normally as possible, for several minutes while motion and force data was collected. Subjects ran in a circle around and through the motion capture area, so motion capture data in one constant direction was recorded. Instructions were given to run in a straight line across the data capture area so the subjects would run across the force plates, but subjects were not told about the force plates to minimize possible stride changes. After the subject had successfully completed at least three trials in which the dominant foot struck the force plates subjects were given verbal instructions on altering running form as they continued to run. As they ran, subjects were instructed to attempt to continue running at the same speed while focusing on "sticking the butt out". These instructions were given to achieve the desired testing condition of increased anterior tilt. When three successful trials of anterior tilt were recorded, subjects were instructed to resume running normally. After subjects had run normally for a short period of time, subjects were instructed to continue to run as normally as possible while trying to "tuck the butt underneath" them, which corresponds with increased posterior tilt. Upon completion of three trials of attempted posterior tilt, the subject's involvement in this study was complete. Subjects then completed further motion trials and specific tasks associated with larger study.

The study protocol was incorporated into a larger, funded study in the laboratory, which compensated subjects to increase participation rates. The data from 17 subjects was analyzed. The averages of values for biomechanical loadings, such as ground reaction force, or angles such as pelvic tilt, for each treatment condition by subject were calculated. Forces and moments were normalized by body weight and height. A regression analysis via a mixed effects regression model was used to examine outcomes after averaging the three trials for each subject and treatment to determine what correlations, if any, were found between pelvic tilt and biomechanical loadings associated with injury risk. To adjust for subject-level differences, a random intercept was included for each subject. The random intercept also helped to account for any correlation associated with repeated measures on the same subject. No adjustments were made for multiple comparisons. All tests were two sided to account for positive or negative changes in outcomes. Inference was performed on the fixed effect of the treatment group. P-values less than 0.05 were considered significant.

c. Population and Sample

This study was a sub-section of a larger study in the OSU Sports Biomechanics Laboratory, which studied former full contact high school football players. Subject recruitment was achieved using a combination of IRB-approved methods – fliers, emailing, and direct verbal communication. Fliers were posted in recreational facilities, restaurants, billboards, community areas, and other areas in which it seemed likely possible subjects may see the fliers. Emails were sent out to groups that included potential subjects, such as recreational sports clubs, fraternities, fitness groups, and personal contacts. Lab members distributed flyers about the study to friends, classmates, and other potential subjects. Inclusion in the larger study required that the subject be a former football player who competed in full contact football no more than five years ago, who also is reasonably fit and active, has not had open abdominal surgery, can comfortably perform a backbend, and could comfortably participate in all required activities of the study. The larger study was funded by a grant from the NFL Charities Foundation, so subjects were compensated for their participation in the study.

Subjects from the larger study selected for this project were determined by data availability. All subjects from the larger study were considered for this project, but subjects for which data was missing or incomplete were excluded. Missing data includes subjects who did not have three full trials in normal running, attempted anterior tilt, and attempted posterior tilt. Subjects were also excluded if more than one marker was missing for the duration of the trials, in order to minimize possible error from the software program's calculations of biomechanical loadings from the marker set. Several subjects were also excluded from this study due to missing information in the calibration trials at the beginning of the testing session for that subject.

Table 1. Demographic Data for Study Participants – 17 Subjects

	Age (years)	Height (mm)	Weight (kg)
Average	20.35	1814.98	86.11
St. Dev.	1.00	55.78	9.58
Min	18	1752.60	74.39
Max	22	1924.05	115.21

d. Data and Instrumentation

Data collection occurred in the Sports Biomechanics Laboratory located in the Martha Morehouse Medical Pavilion, which is part of the Ohio State University Medical Center. The laboratory occupies half of what used to be a gymnasium, or about 3200 square feet, so the data capture area is fairly large. Eight high speed Vicon MX-40 infrared cameras and four Bertec 4060-10 force plates embedded in the floor capture movement and force data. The camera system was calibrated at the beginning of each day to minimize camera and capture errors. During testing, the infrared cameras detect the reflection of light from the markers as the subject moves through the data capture area. The Vicon/Nexus software system worked with the camera system to create a virtual three-dimensional representation of the capture area and the subject.

Data processing and analysis occurs using Nexus and MatLab software. Data processing is a multi-step, time-intensive process. The software program that captured the motion of the reflective markers creates a three-dimensional virtual skeleton of the markers only. The markers are labeled with anatomical names so the software can make a better model of the motion. The trajectories, or motion, of the markers must be checked and corrected to remove any gaps or errors in the movement of the markers. After labeling and filling gaps have been completed, MatLab programs written by lab members are run to create the final virtual skeleton of the processed subject data. The large number of markers used during motion capture is used to account for soft tissue movement, and the MatLab programs use the motion of the markers as well as anatomical measurements to determine the actual movement of the subject's skeleton during the motion trials.

The Nexus software is used in combination with MatLab software and special coding programs developed by Dr. Ajit Chaudhari to interpret the data collected by the motion capture system. The software program analyzes the motion of the markers to calculate biomechanical loadings, pelvic angle, trunk position, and joint angles. For this study, pelvic tilt was defined as the tilt of the pelvis relative to horizontal, rather than tilt relative to the femora or the spine. After the software program calculates these values, values of interest can be exported into an Excel worksheet. The data allows investigators to quantify the forces that occur as the subject moves.

Chapter 3

a. Results

When the anterior and posterior pelvic trials were compared to the control trials, the dependent variables of peak tibial rotation, hip adduction angle, hip adduction moment, knee adduction moment, mean knee adduction moment, vertical ground reaction force, knee adduction impulse, hip adduction impulse, speed and pelvic tilt were considered. The regression analysis found that subjects were able to alter pelvic tilt in a significant manner. Compared to control trials, subjects were able to tilt their pelvis, on average, 4.56 degrees more in the anterior direction (P=0.0001) and 3.62 degrees more in the posterior direction (P=0.0008). The analysis also found that very few loadings were significantly changed when subjects were attempting anterior or posterior tilt. The only loadings that changed with significance were hip adduction moment and hip adduction impulse when comparing control to attempted anterior tilt trials. Higher peak hip adduction moment and hip adduction impulse have been previously associated with iliotibial band syndrome (ITBS) (16). Both loadings were reduced when comparing anterior tilt to the control trials, which may mean that this alteration of running form reduces ITBS injury risk. No other loadings were found to be significant.

Table 2. Least Squares Mean Differences

	Trial Type	Mean Difference	P Value	95% Confidence Interval
Hip Adduction Moment (%BW-ht)	C vs. A	0.8499	0.0383	0.0486, 1.651
Hip Adduction Impulse (%BW-ht-s)	C vs. A	0.1039	0.0395	0.0053, 0.2025
Anterior Tilt (deg)	C vs. A	-4.5647	<0.0001	-6.1882, -2.9412
Posterior Tilt (deg)	C vs. P	3.6244	0.0008	1.752, 5.4968

For C vs. A trials, positive value indicates that the control trial had a greater value by that mean difference. The negative value for control vs. anterior trial types means that the subjects tilted their pelvis further by 4.5 degrees in the anterior direction. The positive value for control vs. posterior trial type indicates that the subjects tilted their pelvis by 3.6 degrees in the posterior direction. Pelvic tilt was calculated relative to horizontal.

Qualitative Observations

Subjects responded to the verbal instructions in a variety of ways. Sometimes subjects seemed to understand what researchers were trying to instruct them to do, and visibly appeared to change pelvic tilt while running. The majority of subjects, however, seemed to lean forward when asked to “stick their butt out” (anterior tilt) or lean backward when asked to “tuck their butt underneath them” (posterior tilt). Subjects appeared to compensate for the instructed changes by changing other aspects of their running form other than just pelvic position. A few subjects, when given instructions for posterior tilt, pulled their shoulders back and thrust their chins up in the air. Though they were asked to continue running at a constant speed, most subjects appeared to change their running speed when attempting to alter their form. Most slowed down, but a few appeared to speed up instead. During visual inspection, subjects did not appear to be able to change pelvic tilt or to have an awareness of other changes in running form during these attempts. The observed inability of subjects to comfortably alter pelvic tilt during motion was confirmed by the very small calculated changes in pelvic tilt, as changes in such small degree would not have been very noticeable to the researchers.

b. Discussion

Subjects were able to change the tilt of their pelvis and alter their running form in a significant way, though the overall change was small. A change in pelvic tilt of a few degrees produced a statistically significant decrease in the hip adduction moment and impulse for when subjects performed anterior tilt trials. These results suggest that increased anterior tilt produces a decrease in biomechanical loadings that may decrease risk of iliotibial band syndrome, as previous research has identified, via prospective study, that runners who later had ITBS had higher hip adduction moment and impulse [13]. Other loadings demonstrated no significant change, which could be due to a variety of reasons.

The changes in hip adduction moment and impulse for anterior tilt trials could be due to a number of factors. Different muscle activation, such as increased usage of lower back muscles or hamstring muscles, could be responsible for the effect. Different muscle activation was not included in the scope of this project but could be tested in the future via electromyography, or EMG, studies. Another possible explanation is that subjects stiffened up their entire hip musculature to achieve the change in pelvic tilt, which can reduce total range of motion. This may have impacted the hip adduction moment and impulse. While muscle activation data is not available for this data set, the results of this pilot study suggest that in the future, it is critical to incorporate measurements of muscle activity in order to understand how subjects are making changes to the loadings. Spine angle and torso lean may play a role in the changes effected by a change in pelvic tilt.

Considering the actual change achieved by subjects reveals that the changes were minimal. Pelvic tilt relative to the horizontal plane changed by a few degrees only, and the change in hip adduction moment and impulse was very small, at 0.85%BW-ht and 0.1%BW-ht-s change in loadings, respectively. These changes, while statistically significant, do not seem to reduce injury risk enough to warrant a permanent change in running form.

Limitations and Future Work

While the subject demographics show that the subjects are very similar to each other, which helps to minimize the effects of other, uncontrollable subject differences, the subject population may have altered the results of the study. Subjects, while generally fit and active, were not “runners” who trained by running. More significant results may have been observed if the study population was more oriented towards running as the primary physical training activity. This could be achieved by recruiting subjects who run a certain number of miles per week or who had a certain number of years of experience with running. Those who run with more frequency or intensity are likely to have better awareness of their running form and may be better able to control changes so that only pelvic tilt is altered.

Significant change in pelvic tilt was observed, but larger effects and greater control may be observed in future studies if subjects have the opportunity to practice these movements. The effect of pelvic tilt on biomechanical loadings may have had a more variable effect because the action was new to the subject and instructions were somewhat open to interpretation. As a long term goal of research such as this is to train people to affect their own lower extremity loadings in a way that injury risk will be reduced, the verbal instruction utilized in this study showed that significant changes can be achieved through changes in training. Simple verbal instructions were sufficient to guide subjects to make changes in running form, but physical demonstration of the desired change in pelvic tilt could increase the efficacy of causing change in running form. Providing practice time so that these alterations of form feel more natural for the subject could produce significant effects in other loadings or increase the ability to change pelvic tilt to a larger degree. Additional controls such as having subjects run on a treadmill to practice the new motion and to control the speed of the subject would be valuable. Another important consideration in changing running form via pelvic tilt is the subjects' ability to tilt the pelvis at rest. The maximal range of motion of tilting the pelvis in the anterior or posterior direction was not tested or recorded, and in future studies, this data should be collected to compare the subjects' ability to tilt the pelvis during movement to overall range of motion.

Further study is recommended. Additional investigation into the effect of pelvic tilt and other alterations of running form on biomechanical loadings seems reasonable. Additional controls and data collection should be utilized, such as testing overall range of motion, employing EMG testing for muscle activation, and consideration of other factors such as stride length. A different study population may also yield different results, so studying the effect of altered running form with a running specific subject subset may be useful. If small pelvic tilt changes can produce significant, albeit small, changes in loadings associated with injury, additional research into this area may lead to better insights and instruction methods for those at risk of overuse injury.

References

1. Andriacchi, T.P., et al., “A point cluster method for in vivo motion analysis: applied to a study of knee kinematics.” *Journal of Biomechanical Engineering*,1998. 120(6): p. 743-9.

2. Chaudhari AM, Hearn BK, Andriacchi TP. "Sport Dependent Variations in Arm Position Influence Knee Loading: Implications for ACL Injury." *American Journal of Sports Medicine*, 2005: 33(6): 824-830.

3. Cole GK, Nigg BM, van den Boger AJ, Gerritsen KGM. "Lower extremity joint loading during impact in running." *Clinical Biomechanics,* 1996: 11(4): 181-193.

4. Fields K, Bloom J, Priebe D, Foreman B. "Basic Biomechanics of the Lower Extremity." *Prim Care Clin Office Pract.*, 2005: 32: 245-251.

5. Grimston SK, Engsberg JR, Kloiber R, Hanley DA. "Bone Mass, External Loads, and Stress Fracture in Female Runners." *International Journal of Sport Biomechanics*, 1991.

6. Hamill J, van Emmerik REA, Heiderscheit BC, Li L. "A dynamic systems approach to lower extremity running injuries." *Clinical Biomechanics*, 1999: 14(5): 297-308.

7. Hewett TE, Myer GD, Ford KR, et al. "Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study." *American Journal of Sports Medicine*, Apr 2005: 33(4): 492-501.

8. Hodges, PW. "Core stability exercise in chronic low back pain." *The Orthopedic Clinics of North America,* Apr 2003: 34(2): 245-254.

9. Jamison, S. "Determining the Correlation between Biomechanical Loads Indicative of Over-Use Running Injuries and Core Strength and Stability." *The Ohio State University Department of Mechanical Engineering - Honors Thesis*, 2008.

10. Kibler WB, Press J, Sciascia A. "The role of core stability in athletic function." *Sports Medicine*, 2006: 36(3): 189-198.

11. Laughton CA, Davis IM, Hamill J. "Effect of strike pattern and orthotic intervention on tibial shock during running." *Journal of Applied Biomechanics,* May 2003: 19(2): 153-168.

12. Leetun DT, Ireland ML, Willson JD, Ballantyne BT, McClay DI. "Core Stability Measures as Risk Factors for Lower Extremity Injury in Athletes." *Medicine and Science in Sports and Exercise*, 2004: 36: 926-934.

13. MacMahon JM, Chaudhari AM, Andriacchi TP. "Biomechanical Injury Predictors for Marathon Runners: Striding Towards Iliotibial Band Syndrome Injury Prevention." *Conference of the International Society of Biomechanics in Sports.* Hong Kong, 2000.

14. Scott SH, Winter DA. "Internal forces of chronic running injury sites." *Medicine and Science in Sports and Exercise*, Jun 1990: 22(3): 357-369.

15. Thomas SS, Supan TJ. "A Comparison of Current Biomechanical Terms." *Journal of American Orthotists and Prosthetists*, 1990: 2(2): 107-114.

16. Wolchok JC, Hull ML, Howell SM. "The Effect of Intersegmental Knee Moments on Patellofemoral Contact Mechanics in Cycling." *Journal of Biomechanics*, 1998: 31(8): 677-683.

17. Zazulak BT, Hewett TE, Reeves NP, Goldberg B, Cholewicki J. "Deficits in Neuromuscular Control of the Trunk Predict Knee Injury Risk: A Prospective Biomechanical-Epidemiologic Study." *American Journal of Sports Medicine*, Jul 2007: 35(7): 1123-1130.

Appendix A

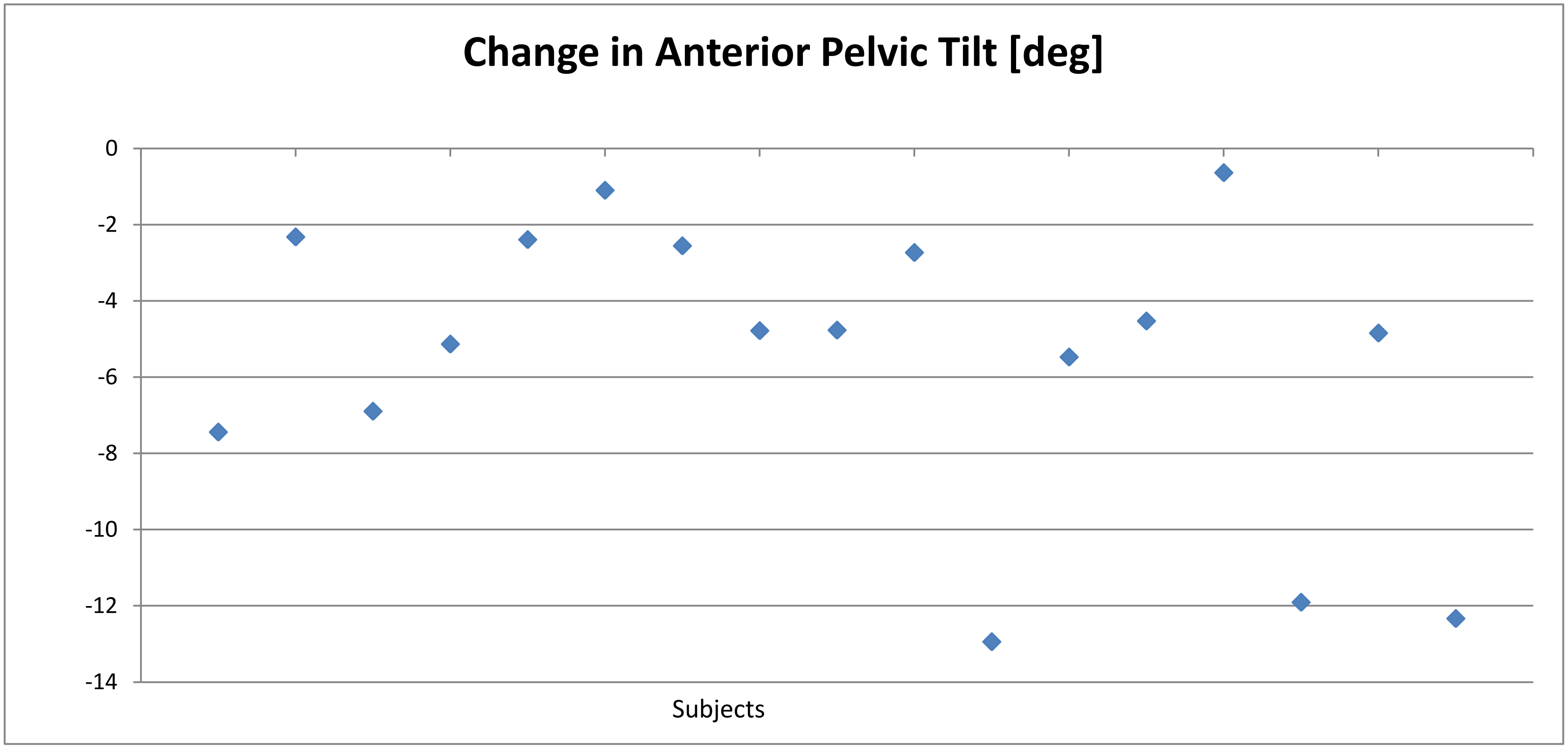

Figure 6. Difference of Averages for Change in Anterior Pelvic Tilt vs. Control

The y-axis represents the change in pelvic tilt, in degrees, for control vs. anterior pelvic tilt. The negative value indicates that the subjects increased pelvic tilt in the anterior direction, relative to horizontal.

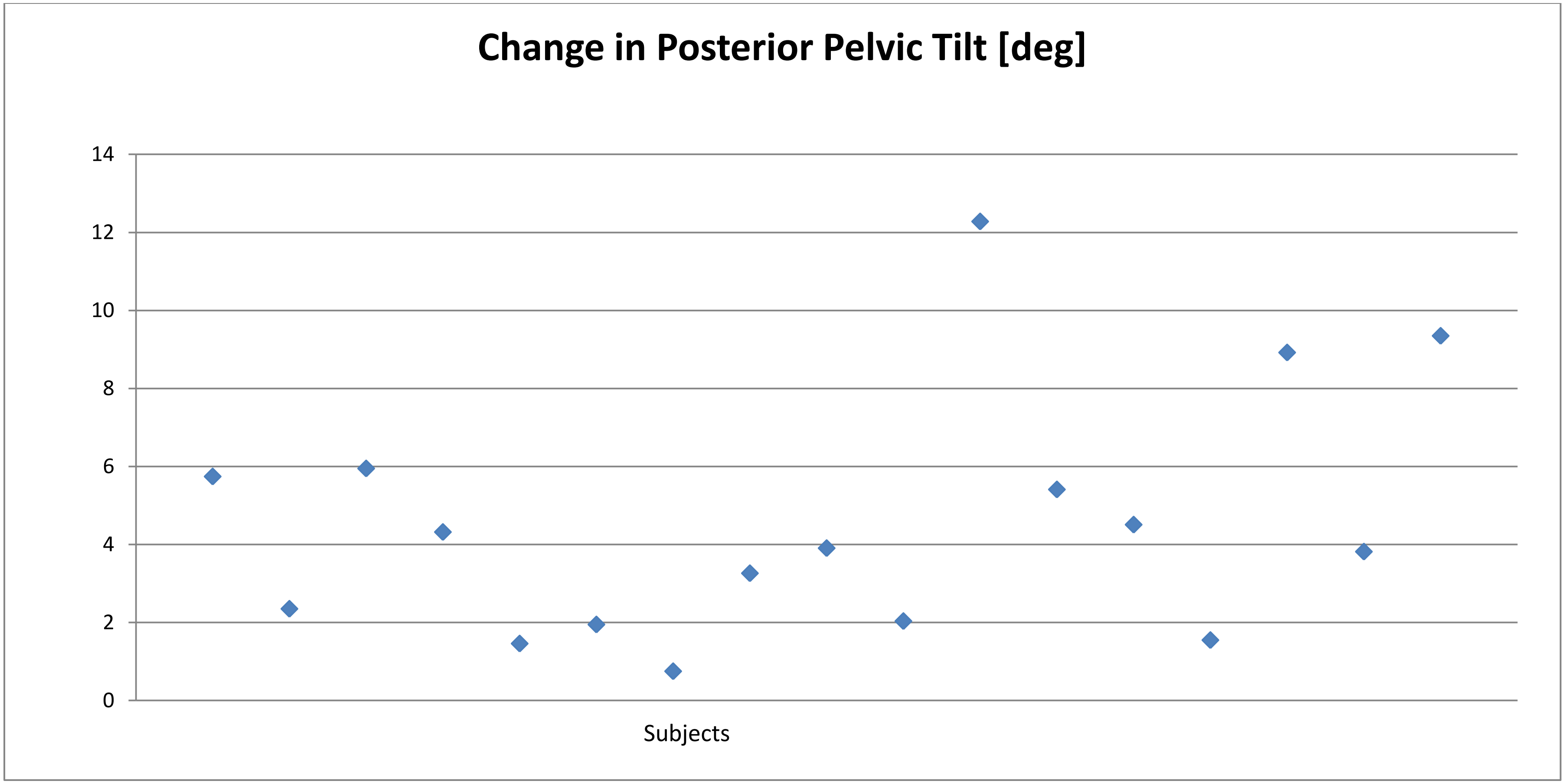

Figure 7. Difference of Averages for Change in Posterior Pelvic Tilt vs. Control

The y-axis represents the change in pelvic tilt, in degrees, for control vs. posterior pelvic tilt. The positive value indicates that the subjects increased pelvic tilt in the posterior direction, relative to horizontal.

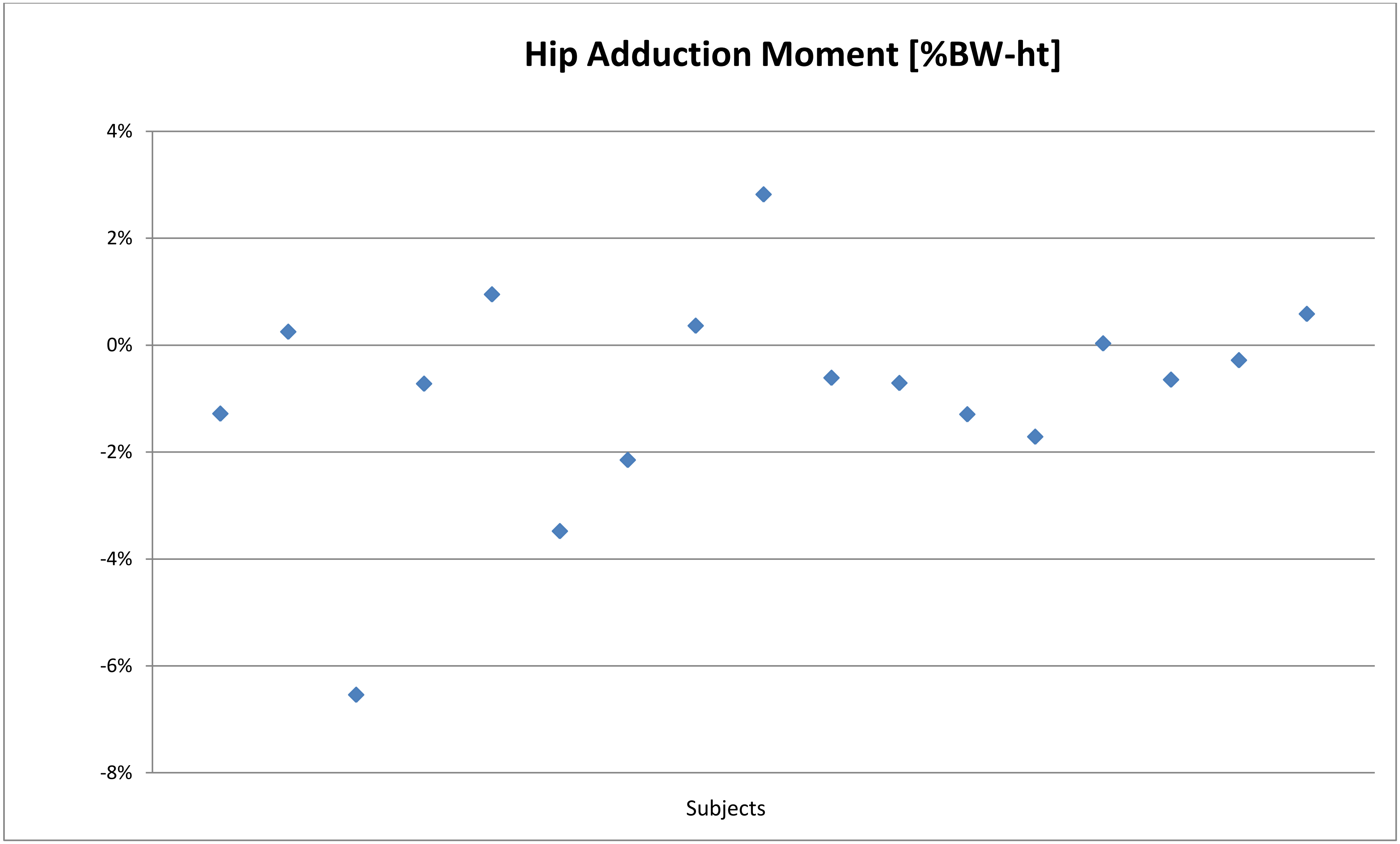

Figure 8. Difference of Averages for Hip Adduction Moment for Anterior Pelvic Tilt vs. Control
Normalized by bodyweight and height, represents absolute change in moment

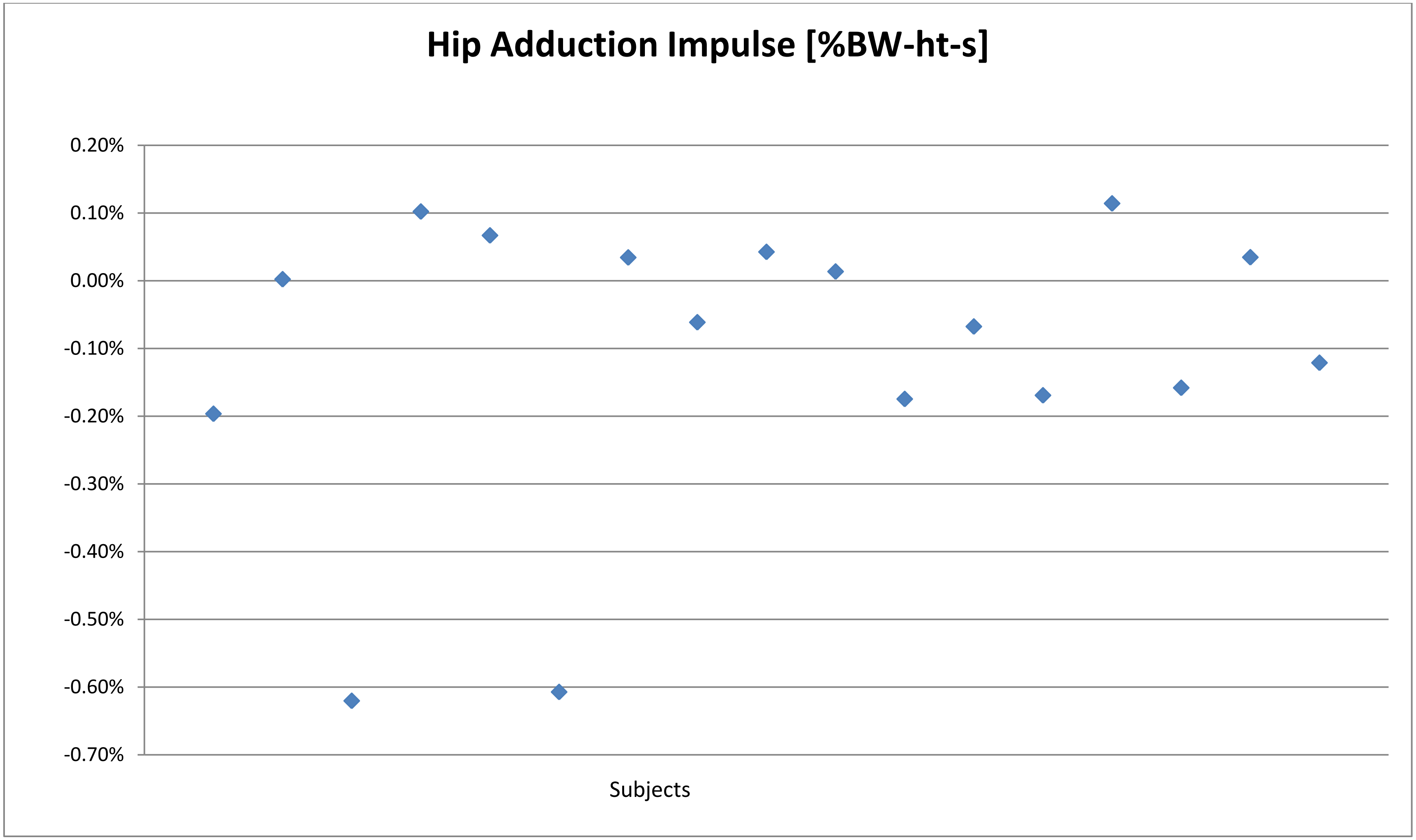

Figure 9. Difference of Averages for Hip Adduction Impulse for Anterior Pelvic Tilt vs. Control Normalized by bodyweight and height

Table 3. Least-squares Mean Differences – Full Data Set

Outcome	Trial type	Trial type	Estimate	Standard Error	DF	t Value	Pr > |t|	Alpha	Lower 95% CI	Upper 95% CI
Hip Adduction Moment (%BW-ht)	P	C	-0.6320	0.3934	32	-1.61	0.1179	0.05	-1.433	0.1691
Hip Adduction Moment (%BW-ht)	P	A	0.2177	0.3934	32	0.55	0.5838	0.05	-0.584	1.019
Hip Adduction Moment (%BW-ht)	C	A	0.8499	0.3934	32	2.16	0.0383	0.05	0.0486	1.651
Knee Adduction Moment (%BW-ht)	P	C	-0.1230	0.1701	32	-0.72	0.4746	0.05	-0.4700	0.2234
Knee Adduction Moment (%BW-ht)	P	A	0.1051	0.1701	32	0.62	0.5409	0.05	-0.2410	0.4517
Knee Adduction Moment (%BW-ht)	C	A	0.2282	0.1701	32	1.34	0.1892	0.05	-0.1180	0.5747
Vertical ground reaction force (%BW)	P	C	0.2352	3.37	32	0.07	0.9448	0.05	-6.629	7.099
Vertical ground reaction force (%BW)	P	A	3.929	3.37	32	1.17	0.2523	0.05	-2.935	10.79
Vertical ground reaction force (%BW)	C	A	3.693	3.37	32	1.1	0.2812	0.05	-3.171	10.56
Knee Adduction Impulse (%BW-ht-s)	P	C	-0.011	0.0272	32	-0.39	0.6999	0.05	-0.066	0.0448
Knee Adduction Impulse (%BW-ht-s)	P	A	0.0208	0.0272	32	0.77	0.4496	0.05	-0.035	0.0762
Knee Adduction Impulse (%BW-ht-s)	C	A	0.0314	0.0272	32	1.15	0.2569	0.05	-0.024	0.0868
Hip Adduction Impulse (%BW-ht-s)	P	C	-0.048	0.0484	32	-0.99	0.3287	0.05	-0.147	0.0506
Hip Adduction Impulse (%BW-ht-s)	P	A	0.0559	0.0484	32	1.15	0.2568	0.05	-0.043	0.1545
Hip Adduction Impulse (%BW-ht-s)	C	A	0.1039	0.0484	32	2.15	0.0395	0.05	0.0053	0.2025
Anterior Tilt (degrees)	C	A	-4.5647	0.7659	16	-5.96	<.0001	0.05	-6.1882	-2.9412
Posterior Tilt (degrees)	P	C	3.6244	0.8832	16	4.1	0.0008	0.05	1.752	5.4968

Data table courtesy of Gregory Young. Significant findings highlighted in yellow.