TRITSCH, AMANDA JO, Ph.D. Associations between Physical Characteristics and Landing Biomechanics in Adolescent Females (2013)
Directed by Dr. Sandra J. Shultz. pp.156

Females tear their anterior cruciate ligament (ACL) at a rate of four to six times that of similarly trained males. This greater risk of ACL injury in females begins to emerge around age 12, then peaks and remains elevated from age 15 into adulthood. While the specific underlying factors that lead to this increased risk in females are yet unknown, this is the time that sex differences in physical characteristics and landing biomechanics begin to emerge. During adolescence, females develop higher risk landing strategies that are thought to place them at greater risk for ACL injury. As females mature, they perform landing maneuvers with greater knee valgus, a more extended knee, and increased reliance on the knee extensor muscles to dissipate landing forces. While these biomechanical patterns are associated with greater strain on the ACL, it is unclear what causes females to develop these higher risk knee biomechanics. The trend towards higher risk landing strategies in adolescent females occurs during a time of steady growth, when strength, body composition, and fitness levels are disparately changing in females and males. Maturing females develop greater fat mass which is associated with a plateau in relative strength to body weight and decreased cardiovascular fitness.

Therefore, the purpose of this study was to examine the extent that physical measures of strength, body composition and fitness affect knee joint biomechanics during a landing task in adolescent females. Physical characteristics were assessed using a battery of field based assessments for cardiorespiratory fitness, musculoskeletal fitness (strength), and body composition that has been demonstrated to be reliable and valid in

children and adolescents. Because both physical measures (e.g. fitness, strength), and knee biomechanics may be affected by fatigue (when knee injury is more likely to occur), the relationship between physical characteristics and knee biomechanics was examined both before and after an exercise challenge.

Fifty adolescent females between the ages of 11 and 15 were used for analyses (Age: 12.7±1.4yrs, Tanner stage: 3.4±0.8, Height: 160.7±7.8 cm, Mass: 52.3±10.2 kg). The primary findings were that a greater Tanner stage of maturation was related to less predicted initial knee valgus angle following exercise (R^2=0.082, p=0.04), while greater functional lower extremity strength was related to greater predicted peak internal tibial rotation angle both before (β=0.18, p=0.01) and after exercise (β=0.17, p=0.03). There were no associations between physical characteristics and relative energy absorption at the knee. Furthermore, exercise had little to no effect on these associations.

These results indicate that the measures of maturation and strength are related to landing biomechanics both before and after exercise. Thus, it appears that landing mechanics that have been shown to change during adolescence cannot solely be attributed to potential changes in strength, body composition and fitness based on the field tests used. However, each of these physical characteristic warrants further inclusion in future studies investigating changing landing biomechanics in populations of adolescent females that participate in athletics. Though only functional strength was statistically related to at-risk landing biomechanics in this representative population of adolescent females, the strength of the relationships with other variables suggests that with more subjects, particularly in Tanner stages 1 and 5, additional relationships may emerge.

ASSOCIATIONS BETWEEN PHYSICAL CHARACTERISTICS AND LANDING BIOMECHANICS IN ADOLESCENT FEMALES

by

Amanda Jo Tritsch

A Dissertation Submitted to
the Faculty of The Graduate School at
The University of North Carolina at Greensboro
in Partial Fulfillment
of the Requirements for the Degree
Doctor of Philosophy

Greensboro
2013

Approved by

_____
Committee Chair

To Jen:

You had to endure the brunt of my frustration over the past four years and never stopped believing I could do it. Thank you for your patience and understanding.

To my family:

For always believing in me – even when I didn't believe in myself. Mom, Dan, Dad, Melissa, Jen, Luke, Darron, and Misha thank you for all you have given me and for always being there for me. I know that not everyone is presented with the opportunities I have been fortunate enough to experience and I hope I never let you down. My grandparents Doris, Myron, Chuck, and Joan who always told me growing up that I could be whatever I wanted and accomplish any goal I set for myself. The memory of my grandmother and all of the encouraging words she shared with me will continue to push me daily both academically and personally.

APPROVAL PAGE

This dissertation written by AMANDA JO TRITSCH has been approved by the following committee of the Faculty of The Graduate School at The University of North Carolina at Greensboro.

Committee Chair _____
Dr. Sandra J. Shultz

Committee Members _____
Dr. Randy J. Schmitz

_____
Dr. Laurie Wideman

_____
Dr. Kelly L. Rulison

_____
Date of Acceptance by Committee

_____
Date of Final Oral Examination

ACKNOWLEDGEMENTS

Dr. Sandra J. Shultz, for her professional dedication and mentorship to me, both personally and professionally, throughout this project and my entire doctoral experience.

My committee, for always being willing to meet with me for advice, as well as so often providing amazing feedback.

My lab colleagues, for always listening while I talked through problems, helping me come to the best solutions, and making coming to work fun.

Funding for this project provided by the National Athletic Trainers' Association Research & Education Foundation - Doctoral Research Grant

TABLE OF CONTENTS

LIST OF TABLES

CHAPTER I

INTRODUCTION

Statement of Problem

Females tear their ACL at a rate of four to six times that of similarly trained males (E. Arendt & Dick, 1995; E. A. Arendt, Agel, & Dick, 1999; Deitch, Starkey, Walters, & Moseley, 2006; Gomez, DeLee, & Farney, 1996). This greater risk of ACL injury in females begins to emerge around age 12 (Gianotti, Marshall, Hume, & Bunt, 2009; Le Gall et al., 2006; Peterson, Junge, Chomiak, Graf-Baumann, & Dvorak, 2000) and then peaks and remains elevated from age 15 into adulthood (Csintalan, Inacio, & Funahashi, 2008; Shea, Pfeiffer, Wang, Curtin, & Apel, 2004). While the specific underlying factors that lead to this increased risk in females are yet unknown, this is the same time that sex differences in physical characteristics and landing biomechanics begin to emerge.

During adolescence, females develop higher risk landing strategies compared to males, which are thought to place them at greater risk for ACL injury (Ford, Myer, & Hewett, 2010; Ford, Shapiro, Myer, Van Den Bogert, & Hewett, 2010; Hass et al., 2003; Hass et al., 2005; Sigward, Pollard, & Powers, 2011; Yu et al., 2005). As females mature, they perform landing maneuvers with greater knee valgus (Ford, Myer, & Hewett, 2003; Hewett, Myer, & Ford, 2004), a more extended knee (Hass, et al., 2003; Hass, et al., 2005; Yu, et al., 2005), and increased reliance on the knee extensor muscles to dissipate landing forces (Sigward, et al., 2011). While cadaveric studies suggest these

biomechanical patterns are associated with greater strain on the ACL (Gabriel, Wong, Woo, Yagi, & Debski, 2004; Sakane et al., 1999; Withrow, Huston, Wojtys, & Ashton-Miller, 2006; Woo et al., 1998), it is still unclear what causes females to develop these higher risk knee biomechanics in the first place.

The trend towards higher risk landing strategies in adolescent females occurs during a time of steady growth, when strength (Buchanan & Vardaxis, 2003), body composition (Loomba-Albrecht & Styne, 2009), and fitness levels (Janz, Dawson, & Mahoney, 2000; McMurray, Harrell, Bangdiwala, & Hu, 2003) are disparately changing between sexes. Maturing females develop greater fat mass (Loomba-Albrecht & Styne, 2009) which is associated with a plateau in relative strength to body weight (Buchanan & Vardaxis, 2003) and decreased cardiovascular fitness (Janz, et al., 2000). Conversely, maturing males increase their relative lean mass and strength while maintaining fitness (Buchanan & Vardaxis, 2003; Janz, et al., 2000; Loomba-Albrecht & Styne, 2009). While many of the changing physical characteristics are related to changes in hormone levels (Oerter, Uriarte, Rose, Barnes, & Cutler, 1990; Rose et al., 1991), they are, to an extent, modifiable with proper training. Authors have suggested that these physical changes, in particular, strength and body composition, may raise the center of mass in females at this time (Ford, Shapiro, et al., 2010), and contribute to the increased knee valgus movement during landing (Hewett, et al., 2004). Others suggest that decreased neuromuscular control about the knee may result from the hamstring strength stasis in females (Ahmad et al., 2006; Barber-Westin, Galloway, Noyes, Corbett, & Walsh, 2005). This is supported by limited research in adult males and females that found increased

BMI (an estimate of body fatness) to be a risk factor for ACL injury (Evans et al., 2012; Uhorchak et al., 2003), while lower extremity strength (R. J. Schmitz & Shultz, 2010) and lean mass (Montgomery, Shultz, Schmitz, Wideman, & Henson, 2012) are related to sex specific landing patterns where increased strength is related to increased energy absorption about the knee in females, but the same relationship is not found in males. However, no research to date has directly examined whether physical characteristics that change during maturation are associated with the changes in knee biomechanics during this time. Moreover, there is a need to examine these physical characteristics using accessible, field based measures so that we can more readily screen for these risk factors in the future (DiStefano, Padua, DiStefano, & Marshall, 2009; Shultz et al., 2010).

Therefore, the purpose of this study was to examine the extent that physical measures of strength, body composition and fitness effect knee joint biomechanics during a landing task in adolescent females. Physical characteristics were assessed using the ALPHA (Assessing Levels of Physical Activity) which is a battery of field based assessments for cardiorespiratory fitness, musculoskeletal fitness (strength), and body composition that has been demonstrated to be reliable and valid in children and adolescents (Ruiz et al., 2011). Because both physical characteristics (e.g. fitness, strength) and knee biomechanics may be affected by fatigue (Borotikar, Newcomer, Koppes, & McLean, 2008; Sanna & O'Connor, 2008) (when knee injury is more likely to occur (Price, Hawkins, Hulse, & Hodson, 2004)), the relationship between physical characteristics and knee biomechanics was examined both before and after an exercise challenge.

Objective and Hypothesis

The objective was to determine the extent to which physical characteristics (that are modifiable through training) predict high risk knee joint landing biomechanics in adolescent females, both before and after an exercise challenge.

Hypothesis 1: One or a combination of greater body fat, decreased strength, and decreased cardiovascular fitness, will predict greater knee valgus angle, greater tibial rotation angle, and greater relative sagittal energy absorption about the knee, throughout the deceleration phase of landing.

Hypothesis 2: Following an exercise challenge, one or a combination of greater body fat, decreased strength, and decreased cardiovascular fitness, will predict greater knee valgus angle, greater tibial rotation angle, and greater relative sagittal energy absorption about the knee, throughout the deceleration phase of landing.

Hypothesis 3: The relationship between the predictor variables (one or a combination of body fat, strength, and cardiovascular fitness) and knee joint biomechanics will be stronger when knee joint biomechanics are measured after the exercise challenge compared to when measured before the exercise challenge.

Limitations and Assumptions

1. Results from this dissertation cannot be generalized to populations other than the adolescent females aged 11 to 15 studied, or to tasks other than the drop jump.
2. All participants provided a maximum effort during testing.
3. Inverse dynamics calculations represent the total moments occurring at the joint.

4. Motion capture frequency of 240 Hz was great enough to accurately track lower extremity joint motions.
5. The motion capture system (Phase Space, San Leandro, CA) can accurately track and identify individual active markers, each with individual frequencies, with high reliability.
6. This work does not account for other anatomical and hormonal risk factors that are potentially associated with high-risk knee joint biomechanics.
7. Kinematics and kinetics of the drop jump were assessed in the laboratory which may be different from the participant's normal playing surface and playing activity.
8. A single tester obtained all skinfold measures; therefore prediction equations may not be generalizable to other testers.

Delimitations

1. Only healthy adolescent females between the ages of 11 and 15 who have had no musculoskeletal injury to either lower extremity in the past 6 months and have no history of surgery on either lower extremity participated in this study.
2. Kinetic and kinematic measurements were only obtained from the left leg.
3. Data, results, and interpretation were based on the deceleration phase of the initial landing of a double legged drop jump maneuver.
4. All athletes wore the same lab issued shoe model as opposed to their personal footwear usually worn during training.

5. The landing maneuver was pre-planned, which may not adequately represent reactive response to another athlete or an implement as can occur in practice or competition.
6. The regression analyses used to represent the models to estimate the dependent variables are only evident of the predictor variables included in them and do not account for other alignment, biomechanical, or neuromuscular factors that may interact with these variables to influence the dependent variables.
7. Skinfold measurements were obtained by a single researcher with established day-to-day reliability.

Operational Definitions

Healthy: No history of injury to either lower extremity in the past 6 months that has limited normal activities; no previous history of injury to the capsule, ligaments, or menisci of either knee; no previous history of surgery to either lower extremity; no vestibular or balance disorders that could cause them to lose their balance during the functional tasks; and no history of cardiovascular disease.

Recreationally active: Current participation in a minimum of 90 minutes of exercise per week and currently active in sport activities that include running and cutting and landing maneuvers such as soccer, basketball, and lacrosse.

Adolescent: 11 to 15 years old.

Modifiable physical characteristics: The physical factors of body composition, strength, and cardiovascular fitness.

Drop Jump: Task that begins in a standardized take-off position with the front edge of participants' shoes aligned along the front edge of the platform and hands placed at ear level; participants are instructed to drop off the platform, without jumping or stepping, and perform a maximal vertical jump upon landing.

Deceleration phase: Period between ground contact (vGRF > 10N) and peak knee flexion of drop jump task.

Predictor Variables

Maturation: Tanner stage of maturation (ranging from 1 (pre pubertal) to 5 (adult level of development)) as determined by self-assessment pubertal development using standardized series of drawings with explanatory text (Leone & Comtois, 2007; K. E. Schmitz et al., 2004).

Body Fat: Percent body fat as calculated using the skinfold thickness of the triceps and subscapular regions and the Slaughter equations of:

$$\text{skinfolds} < 35 \text{ mm} = 1.33(\text{sum of 2 skinfolds}) - 0.013(\text{sum of 2 skinfolds})^2 - 2.5$$

$$\text{skinfolds} > 35 \text{ mm} = 0.546(\text{sum of 2 skinfolds}) + 9.7$$

Standing Broad Jump: Maximum jumping distance was a measure of lower extremity muscle strength. Participants began standing on both feet at a start position and were instructed to jump forward as far as possible. Distance from the start position to the back of the heel closest to the start position was measured and reported in centimeters (cm).

Cardiovascular fitness: Estimate of VO_{2max} was calculated via performance (distance run) on the 20 meter shuttle run test, using the Artificial Neural Network (ANN)

equation, which takes into account: sex, age, weight, height, and distance run (Ruiz et al., 2008):

VO_{2max} (ml/(kg min)) =

(1/(1+exp(-(1/(1+exp(-((A1*0.8+(-0.7))*(-1.03329) +
(B1*0.114285714286+(-1.38571428571))*0.54719 +
(C1*0.012213740458+ (-0.406870229008)) * 0.61542 +
(D1*0.0195598978221+(-2.76356892177))*(-0.51381) +
(E1*0.0842105263158+(-0.0684210526316))*(-0.92239) + (-0.34242))))
*(-0.95905)+1/(1+exp(-((A1*0.8+(-0.7))*(-1.19367) +
(B1*0.114285714286+(-1.38571428571))*(-1.54924) +
(C1*0.012213740458+(-0.406870229008))*(-3.18931)+
(D1*0.0195598978221+(-2.76356892177))*0.77773+
(E1*0.0842105263158+(-0.0684210526316))*3.31887+ (-0.55696)))) *
2.19501+1/(1+exp(-((A1*0.8+(-0.7))*1.38191+(B1*0.114285714286 + (-
1.38571428571)) * (-2.14449)+(C1*0.012213740458 + (-
0.406870229008)) *0.0485+(D1*0.0195598978221+(-2.76356892177))
*0.10879+(E1 *0.0842105263158+(-0.0684210526316)) *(-
4.90052)+0.53905))) *(-2.567)+(-0.05105))))-(-0.478945173945))
/0.0204587840012

A1 = sex (boys = 1; girls = 2); B1 = age (year, age range 12 -19 years); C1 = weight (kg); D1 = height (cm); E1 = stage (0.5)

Dependent Variables

Knee Valgus: Abduction angle of the tibia relative to the femur during landing at initial ground contact, peak displacement, and excursion (peak – initial).

Tibial Rotation: Rotation angle of the tibia relative to the femur during landing at initial ground contact, peak displacement, and excursion (peak – initial).

Sagittal Energy Absorption about the Knee: Calculated as the integration of the negative portion of the joint power curve (the product of the normalized joint moment

and joint angular velocity at each time point), and normalized to body weight and height (Joules x BW^{-1} x Ht^{-1}).

Relative Energy Absorption about the Knee: Calculated as the percentage of knee work to total work (hip work + ankle work + knee work).

CHAPTER II

REVIEW OF LITERATURE

This review will address how sex differences in modifiable physical characteristics that emerge during maturation may contribute to high risk knee joint biomechanics and increased risk of ACL injury in adolescent and adult females. Specifically, this review will address the current understanding of ACL injury epidemiology and mechanisms, sex differences in landing biomechanics that have been observed during adolescence and adulthood and are thought to place females at a greater risk for ACL trauma, and how sex differences in physical characteristics that emerge during adolescence may contribute to a female's higher risk landing biomechanics as they mature.

ACL Injury

Injury to the ACL has been described as the "largest single problem in orthopedic sport medicine" (Renstrom et al., 2008). This is in part due to the critical function of the ACL in maintaining knee joint stability as well as the long term ramifications on joint health after sustaining a complete rupture.

Injury and Long-term Implications

The articulating surface of the tibia and femur allows for motion with six degrees of freedom. Meaning, while the greatest amount of rotation occurs in the sagittal plane (flexion and extension), there is also rotation in the frontal plane (creating valgus and

varus angles), and in the transverse plane (internal and external rotation). Additionally, translations occur in all three planes resulting in six types of motion that can occur at the joint. The ACL serves to prevent anterior displacement of the tibia relative to the femur (Butler, Grood, Noyes, & Zernicke, 1978), and protect against excessive tibial rotation and valgus motions and forces (Markolf et al., 1995). Because of the stabilizing role of the ACL, a complete rupture of the ligament often results in a significant joint dysfunction, often requiring reconstructive surgery and six to eight months of rehabilitation. The population that requires surgery has been termed "non-copers" and is the most common subset of ACL deficient individuals (Roewer, Di Stasi, & Snyder-Mackler, 2011; Rudolph, Axe, Buchanan, Scholz, & Snyder-Mackler, 2001)

While the acute trauma of an ACL injury is worrisome, the long-term effects are also cause for concern. For example, it is reported that about 80% of female soccer players show radiographic evidence of osteoarthritis development within 12 years of injury (Lohmander, Ostenberg, Englund, & Roos, 2004). When examined prospectively, those who suffered a knee injury during adolescence and early adulthood had a 3-fold increase in relative risk of developing osteoarthritis by age 65 (Gelber et al., 2000). Aside from the long-term physical costs, medical costs associated with management of ACL injuries in the United States are currently estimated to be $4 billion annually (Brophy, Wright, & Matava, 2009). Collectively these statistics highlight the importance of preventing the initial injury and the subsequent negative effects that can last a lifetime.

Injury Rates Differ Between Males and Females

More than one third of all ACL injuries in the United States each year are suffered by female high school and college athletes, despite this group representing less than one percent of the total population (Henry & Kaeding, 2001). From adolescence into adulthood, females suffer ACL injuries at a rate of 4-6 times greater than males in sports such as basketball and soccer (E. Arendt & Dick, 1995; E. A. Arendt, et al., 1999; Deitch, et al., 2006; Gomez, et al., 1996; Ireland, 1999). In the largest and longest running epidemiology study to date, the ACL injury rate for female versus male NCAA soccer players has been reported to be 0.31 versus 0.13 injuries per 1000 athlete-exposures, where athlete-exposure is defined as one athlete participating in one practice or game where he or she is exposed to the possibility of an athletic injury (E. Arendt & Dick, 1995). Practically speaking, this injury rate translates to one ACL injury in every 385 activity sessions for men, and one in every 161 activity sessions for women (E. Arendt & Dick, 1995).

These sex disparate injury rates are not confined to adults. Sex differences in ACL injury rates also exist in younger age groups (Shea, et al., 2004) with the risk increasing disproportionally in females after age 12 (Gianotti, et al., 2009; Le Gall, et al., 2006; Peterson, et al., 2000), and peaking around age 15 (Csintalan, et al., 2008; Shea, et al., 2004). Based on a national injury registry, similar sex differences in injury rates have been reported for adolescents under age 18 as compared to adults (Parkkari, Pasanen, Mattila, Kannus, & Rimpela, 2008), suggesting this sex difference is maintained into adulthood. When risk was adjusted according to the amount of sports participation and

lifestyle (body composition, smoking history, socioeconomic status, etc), hazard ratios reached 8.5 for females and 4.0 for males. This indicates active females and males were at 8.5 times and 4.0 times greater risk of sustaining an ACL injury, respectively, than the rest of the population (Parkkari, et al., 2008).

Summary

ACL injury and the subsequent joint health problems that result from the injury are accepted as possibly the largest health problem in orthopedic sports medicine (Renstrom, et al., 2008). Moreover, females suffer this injury at a greater rate than males (E. Arendt & Dick, 1995; E. A. Arendt, et al., 1999; Deitch, et al., 2006; Gomez, et al., 1996), with the sex disparity beginning around age 12, (Gianotti, et al., 2009; Le Gall, et al., 2006; Peterson, et al., 2000) peaking around age 15, and remaining higher in females through adulthood (Csintalan, et al., 2008; Shea, et al., 2004). As such, identifying and addressing possible risk factors for injury in an adolescent population, when this rate difference begins to emerge, may be a key factor in preventing this injury.

Injury Mechanism

Injury to the ACL commonly occurs in sporting activities where an athlete runs forward and performs a jump stop maneuver or quickly decelerates or suddenly changes direction (E. Arendt & Dick, 1995), typically on a single leg (Koga et al., 2010; Krosshaug et al., 2007; Olsen, Myklebust, Engebretsen, & Bahr, 2004). Although these deceleration activities are common to many sports, these movement patterns result in injury in some athletes and not in others, and in females more so than males. Despite extensive research, it is still not known specifically when the ACL tears during these

motions, or if the same positions and loads will result in injury for every person. The following section will highlight what is known about the injury mechanisms and the externally applied loads that are known to strain the ACL.

Video-based Analyses

While the specific cause(s) of non-contact ACL injury have yet to be identified, videographic evidence indicates decreased knee flexion, increased knee valgus, and decreased ankle plantar flexion as common positions at landing when ACL injury is thought to occur (Boden, Dean, Feagin, & Garrett, 2000; Boden, Torg, Knowles, & Hewett, 2009; Krosshaug, Nakamae, et al., 2007). In an attempt to describe joint kinematics of actual injury events, six experts with experience in visual analysis of injuries examined videos capturing 30 non-contact ACL injuries in basketball players (Krosshaug, Nakamae, et al., 2007). The authors identified shallow knee and hip flexion at contact in male and female athletes who sustained an injury, with both sexes flexing the knee 15° or less, flexing the hip 30° or less, and landing in 4° or less of knee valgus. However, within a frame of video (33 or 50 ms), hip and knee flexion had doubled in both sexes, and knee valgus doubled in females only (Krosshaug, Nakamae, et al., 2007), which was theorized by the authors to be the result of different knee-loading patterns between men and women upon ACL failure. Video analysis of injury events in female team handball athletes similarly revealed shallow knee flexion angles of 25° or less, but with knee valgus and tibial rotation (relative to the femur), ranging widely from 5-20°, and -15° to 10°, at ground contact, respectively (Olsen, et al., 2004). Whiles these video based studies suggest that ACL injuries often occur in a more upright position, with some

valgus and either internal or external tibial rotation, they suffer from several limitations. Investigators were not able to identify the moment of ACL injury, and in some cases different investigators disagreed by 50 ms on the moment of injury (which may simply reflect the resolution of the video sampling). Further, a follow up study to evaluate the reliability of the video analysis (Krosshaug, Slauterbeck, Engebretsen, & Bahr, 2007) revealed that hip and knee flexion angles were often underestimated, which suggests that the absolute valgus and rotation angles reported in prior work (Krosshaug, Nakamae, et al., 2007) should be interpreted with caution.

Using a more sophisticated model-based image-matching (MBIM) technique, Koga and colleagues (Koga, et al., 2010) were able to further discern knee joint kinematics during actual ACL injury situations by analyzing video broadcasts of 10 additional female athletes suffering ACL tears. This technique built on prior studies by using multiple camera angles to obtain information in all three dimensions, and matching each frame of video to a custom model that reflected the anthropometry of each athlete (Krosshaug & Bahr, 2005). This model includes 21 segments and 57 degrees of freedom, allowing researchers to analyze joint motions throughout the entire body. All injuries analyzed using this technique occurred during either a cutting task or a single leg landing. The authors identified a position of a relatively extended (mean 23°: range 11° to 30°), neutrally abducted (mean 0°: range -2° to 3°), and externally rotated (mean 5°: range -5° to 12°) knee at ground contact. Then, within an average of 40 ms of ground contact, the knee flexed 24°, abducted 12°, and internally rotated 8°, which was then followed by 17° of external rotation from 40-300ms (Koga, et al., 2010). Based on these findings, the

authors suggested that the injury event occurred during the first 40 ms of initial contact (Koga, et al., 2010) and that the subsequent external rotation may result from the ACL rupture, rather than contribute to the ACL rupture as previously thought (Krosshaug, Nakamae, et al., 2007; Olsen, et al., 2004).

Collectively, these studies indicate a knee position near or at the time of ACL injury to reflect a combination of knee extension, valgus, and rotation with valgus and rotation likely resulting from, rather than precipitating the injury. However, these studies are limited to video based observations, and which motions precede versus follow the injury, or pose the greatest strain on the ACL cannot be fully determined from these studies. Subsequently, other studies have examined ACL loading by applying *in vivo* and *in vitro* external loads to the knee to further discern the direction and magnitude of forces that have the potential to stress or injure the ACL.

External Load Application

External loads can be applied to tissue both *in vivo* (in a living organism) and *in vitro*, which uses isolated tissue such as a cadaveric knee. The advantage to these measurement techniques is the ability to directly measure the effects of either isolated or combined loads applied to ACL, potentially shedding more light on specific loading patterns that damage the ACL.

Using a method of applying loads to an intact cadaveric knee, then cutting the ACL and applying loads in the same path, the magnitude and direction of force developed in the ACL can be calculated by subtracting the resultant force of the ACL deficient knee from the ACL intact knee (Sakane, et al., 1999; Woo, et al., 1998).

Studies that have utilized this method have demonstrated that when an anterior load is applied to the tibia with the knee relatively extended (less than 30°), the force within the ACL reflects more than 80% of the applied anterior load (Sakane, et al., 1999; Woo, et al., 1998). This indicates that the ACL is the main passive restraint against anterior directed loads applied to the tibia near full knee extension, which supports the potential negative effects of landing in an erect position.

As tibial rotation has also been implicated as a possible contributing factor to ACL injury (Olsen, et al., 2004), the effect of rotational loads on ACL strain in young healthy patients has been examined by applying a transducer to the ACL during arthroscopic surgery while under local anesthesia (Fleming et al., 2001). Application of a 10 Nm internal rotation moment increased ACL strain by about 3%, however 10 Nm may not represent the physiological values experienced during landing (Fleming, et al., 2001). Physiological values have been found to peak at an average of .34 Nm/kg (Souza et al., 2012) which would translate to 20.4 Nm for a 60 kg person. Cadaveric studies done at 20° of knee flexion demonstrate that internal rotation loads of 31 ± 9.4 Nm result not only in internal rotation (45 ± 18°), but also valgus rotation (11 ± 6.0°) and anterior translation (9.0 ± 3.3 mm) of the tibia relative to the femur (Meyer & Haut, 2008). These combined rotary and valgus loads at both 15° and 30° of knee flexion are reported to significantly increase the tensile force of the ACL (Gabriel, et al., 2004; Kanamori et al., 2000), further implicating the potential for multi-planar motion to strain the ACL.

In an effort to understand the impact of valgus loading during a jump landing, *in vitro* ACL strain under an impulsive axial load (designed to simulate a jump landing) was

examined with and without a valgus knee moment. Specifically, the relative strain difference between a “neutral” loading and a valgus loading with pre-activated knee muscle forces was measured (Withrow, et al., 2006). The valgus load was applied by positioning the knee in 15° of abduction. The addition of the valgus load resulted in 38% greater ACL strain than the “neutral” load (Withrow, et al., 2006). These findings suggest that valgus positioning alone at landing is potentially harmful to the ACL, as compared to a neutral alignment. However, when this protocol was taken a step further by including various combinations of varus-valgus and internal-external rotational loading, peak ACL strain was more sensitive to the direction of axial tibial torque than to the direction of the frontal plane moment (Oh, Lipps, Ashton-Miller, & Wojtys, 2012). Specifically, the mean peak ACL strain under internal tibial torque was 192% greater than under external torque regardless of whether it was coupled with a valgus or varus moment. Thus, the authors suggest that knee valgus loading has a second order of effect on ACL strain that can be explained by the inherent coupling between internal tibial rotation and knee valgus angulation (Oh, et al., 2012). They further suggest that internal tibial rotation induced by an internal tibial torque plays a primary role in increasing the ACL strain and that a knee valgus moment increases the ACL strain by increasing internal tibial rotation (Oh, et al., 2012). Collectively, these studies suggest that while a valgus load increases the strain on the ACL compared to neutral, the load that poses the greatest strain on the ACL is primarily an internal tibial torque. However, this does not preclude valgus motion and torque from being a factor in ACL strain and injury.

Summary

Videographic evaluation of ACL injury suggests that the positioning of the lower extremity within 100 ms of ground contact is critically important to the ACL injury mechanism. The combination of knee valgus in conjunction with tibial rotation during the deceleration phase of landing has been referred to as dynamic valgus, or valgus collapse (Krosshaug, Nakamae, et al., 2007; Olsen, et al., 2004) and is considered a high risk position for ACL injury. Additionally, results from *in vivo* and *in vitro* load application studies indicate that landing in a more erect position strains the ACL (Sakane, et al., 1999; Woo, et al., 1998), and when combined with knee valgus (Gabriel, et al., 2004; Withrow, et al., 2006) and rotation (Fleming et al., 2003; Meyer & Haut, 2008; Oh, et al., 2012) can strain the ACL further. Together this implicates dynamic valgus positioning as a higher risk position for ACL injury. As will be noted in the section to follow, this high risk position is more commonly observed in females.

Sex Differences in Landing Biomechanics

The disparate injury rate between the sexes has led many researchers to examine sex differences in landing biomechanics as a way of potentially pin pointing the specific mechanism that increases ACL injury risk in the female population (Chappell et al., 2005; Huston, Vibert, Ashton-Miller, & Wojtys, 2001; James, Sizer, Starch, Lockhart, & Slauterbeck, 2004; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007; Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; McLean, Huang, Su, & Van Den Bogert, 2004; McLean, Huang, & van den Bogert, 2005; McLean, Lipfert, & Van Den Bogert, 2004; McLean, Neal, Myers, & Walters, 1999; Padua et al., 2009; Pollard,

Sigward, & Powers, 2010; Quatman, Ford, Myer, & Hewett, 2006; R. J. Schmitz & Shultz, 2010; R. J. Schmitz, Shultz, & Nguyen, 2009; Shultz, Nguyen, Leonard, & Schmitz, 2009; Sigward, et al., 2011; Sigward & Powers, 2006; Swartz, Decoster, Russell, & Croce, 2005; Yu, et al., 2005). This section of the review will summarize the findings of kinematic and kinetic differences that have been reported between the sexes in the adult population, and potential theories for the underlying physical differences that may drive these sex differences. A subsequent section will further discuss these underlying theories by examining changes in physical characteristics and knee joint biomechanics that concomitantly occur during maturation.

Kinematic Differences

As previously discussed, a more erect position at ground contact is commonly observed at the time of non-contact ACL injury (Koga, et al., 2010; Krosshaug, Nakamae, et al., 2007; Olsen, et al., 2004). Decreased knee flexion is thought to put the ACL in a less protected position as the patellar tendon/tibial shaft angle is the largest in this range (Grood, Suntay, Noyes, & Butler, 1984; Li, Rudy, Allen, Sakane, & Woo, 1998), which effectively increases the anterior shear forces increasing ligamentous strain (More et al., 1993; Pandy & Shelburne, 1997; Yasuda & Sasaki, 1987) and decreases the ability of the hamstring muscles to produce a counteractive posterior shear force (Pandy & Shelburne, 1997). As such, multiple studies have examined sex differences in knee flexion angle during athletic tasks. Findings among studies are quite varied, with some reporting that females land from a drop jump with less initial knee flexion than males

(Decker, Torry, Wyland, Sterett, & Steadman, 2003; Huston, et al., 2001), while others report no sex difference (Cowling & Steele, 2001; Shultz, et al., 2009).

With the increasing number of non-contact ACL injuries being attributed to twisting and cutting maneuvers, others have examined knee flexion angle during these tasks (Cowling & Steele, 2001; Huston, et al., 2001; Malinzak, et al., 2001; McLean, et al., 1999). During the movements of running and side cutting, maximum knee flexion was reported to occur 10% later in the stance phase for males than females (McLean, et al., 1999), which the authors hypothesized allowed males more time to control and stabilize joint motion in the sagittal plane (McLean, et al., 1999). Examination of a cross-cut maneuver revealed that females flexed their knees 8° less than males through the entire movement of cross-cutting (Malinzak, et al., 2001). Though the timing of knee flexion differed during running and side cutting maneuvers, knee flexion angle was always less in females compared to the males regardless of task (Malinzak, et al., 2001). However, others report no sex differences during similar cutting maneuvers (Cowling & Steele, 2001).

While the varied results and tasks make inference of global movement pattern differences in males vs. females difficult, the preponderance of evidence suggest that females may perform certain tasks with less knee flexion than males (Decker, et al., 2003; Huston, et al., 2001; Malinzak, et al., 2001; McLean, et al., 1999). Lack of consensus with other studies may be attributed to differing task demands between the sexes, such as regulating the speed of running (McLean, et al., 1999) and the height of the drop jump (Decker, et al., 2003; Huston, et al., 2001; Shultz, et al., 2009), which make

the task relatively more difficult for females than males. For example, in the two studies examining landing from a height of greater than 45 cm, females were found to land in a more erect position (Decker, et al., 2003; Huston, et al., 2001). Thus, females may land in a more at-risk position when performing sport related activities that are relatively more demanding for the individual.

Frontal and Transverse Plane Kinematics

Because of the increased strain on the ACL with tibial rotation and valgus loads and motions (Gabriel, et al., 2004; Oh, et al., 2012; Sakane, et al., 1999; Withrow, et al., 2006; Woo, et al., 1998), evidence of these motions during ACL injury events (Koga, et al., 2010; Krosshaug, Nakamae, et al., 2007; Krosshaug, Slauterbeck, et al., 2007; Olsen, et al., 2004), and the general acceptance of dynamic valgus as an at-risk position for injury (Griffin et al., 2006; Krosshaug, Nakamae, et al., 2007; Olsen, et al., 2004; Renstrom, et al., 2008; Shultz, Schmitz, Nguyen, Chaudhari, et al., 2010), investigators have also examined sex differences in frontal and transverse plane kinematics during deceleration type tasks. Among these studies, females are often observed to move with greater knee valgus, or a combination of greater knee valgus and tibial rotation variability (Gehring, Melnyk, & Gollhofer, 2009; Hughes, Watkins, & Owen, 2008, 2010; Kernozek, Torry, H, Cowley, & Tanner, 2005; Malinzak, et al., 2001; McLean, Lipfert, et al., 2004; Walsh, Waters, & Kersting, 2007) compared to males. For example, during cross-cuts, side-cuts, and forward running, females demonstrated 11° greater knee valgus positioning throughout the movements, compared to males (Malinzak, et al., 2001). Sex differences in knee valgus have also been found during landing tasks, with greater knee

abduction angle repeatedly reported in adult females (Gehring, et al., 2009; Hughes, et al., 2008, 2010; Kernozek, et al., 2005; Walsh, et al., 2007). During the stance phase of sidestep cutting, women showed greater variability in knee rotation, with an inter-trial standard deviation (mean ± SD) of 4.8 ± 1.6 in females vs. 2.8 ± 0.9 in males, while men exhibited more variability in hip rotations (4.8 ± 1.8 vs. 2.8 ± 0.8) (McLean, Lipfert, et al., 2004). The addition of a simulated defender does not appear to alter these sex dependent patterns (McLean, Lipfert, et al., 2004). Together, the increased knee valgus and tibial rotation variability commonly seen in female athletes has the potential to place the ACL under greater strain during landing, based on evidence derived from *in vivo* and *in vitro* studies (Gabriel, et al., 2004; Sakane, et al., 1999; Withrow, et al., 2006; Woo, et al., 1998).

Energetics

More recently, energy absorption strategies, which are commonly attributed to eccentric muscle function and the ability to dissipate landing forces, have also been investigated in males and females as it may relate to their potential for ACL injury. Energy absorption is quantified by integrating the negative portion of a power curve (McNitt-Gray, 1993). As power is the product of moment and angular velocity, a negative power curve is generated when the internal moment is opposite to the direction of joint motion. For the deceleration phase of a landing task, this can functionally be interpreted as work done on extensor muscles by eccentric muscle action (Winter, 2009). Current convention suggests absorbing energy through active structures, such as muscles and tendons, is more advantageous than absorbing energy through passive structures,

such as ligaments and bone (Dufek & Bates, 1990; Horita, Komi, Nicol, & Kyrolainen, 1996; James, Dufek, & Bates, 2006; Zhang, Bates, & Dufek, 2000). Moreover, absorbing energy in the larger muscle groups surrounding the hip is considered safer than absorbing energy about the knee (Pollard, et al., 2010; Sigward, et al., 2011).

One of the first studies to examine sex differences in energy absorption during landing found that females absorbed more total energy at the ankle, knee, and hip than males during the impact phase (Decker, et al., 2003). While both sexes absorbed most of the work about the knee, females relied more on the ankle as a secondary absorber, while males relied more on the hip. The authors postulated that this was a result of females landing in a more erect position, thus placing greater demands on the ankle plantarflexors (Decker, et al., 2003). However, this investigation was limited to the energy absorption occurring during the initial impact phase, or first 100 ms of landing. For this reason direct comparison to subsequent studies that examine the entire deceleration phase of landing are somewhat difficult.

Contrary to the aforementioned study, a comparison of energetics when performing a drop jump from a 45 cm box reported that women absorbed 69% more energy about the knee than men, and that both sexes absorbed the greatest amount of energy about the ankle (R. J. Schmitz & Shultz, 2010). The authors attributed this to the relatively more demanding nature of the task for females, thus causing the females to rely more heavily on their knee extensors. This supposition is supported by others who report increased energy absorption about the knee as task difficulty increases (Zhang, et al., 2000). This sex difference may result from sex differences in available lean body mass to

absorb the energy at landing. Specifically, Montgomery, et al. found that greater lean mass predicted more energy absorption at the knee joint, but only in females (Montgomery, et al., 2012). Interpolation of their data indicates that increasing a female's lower extremity lean mass (relative to body mass) by 2% resulted in her energy absorption capabilities about the knee to be the same magnitude as a male's (Montgomery, et al., 2012).

Summary

When landing from a jump, or when cutting or suddenly stopping, adult females and males have been shown to demonstrate a variety of biomechanical differences. These differences have included females landing with less knee flexion (Decker, et al., 2003; Huston, et al., 2001; Malinzak, et al., 2001; McLean, et al., 1999), increased knee valgus (Hughes, et al., 2008, 2010; Kernozek, et al., 2005; Malinzak, et al., 2001; McLean, Lipfert, et al., 2004), greater variability in valgus and rotation (McLean, Lipfert, et al., 2004; McLean, et al., 1999), and energy absorption strategies that rely more heavily on the musculature surrounding the knee (Decker, et al., 2003; R. J. Schmitz & Shultz, 2010). Based on findings of prior sections of this review examining ACL strain behavior with externally applied loads, these landing strategies suggest that females land and change direction in positions that are considered more at-risk for straining the ACL. Some theories put forward for these differences are increased body mass (Hewett, Myer, & Ford, 2006), differences in neuromuscular control (Hewett, et al., 2004), decreased relative strength (Myer et al., 2009), and muscle fatigue (Chappell, et al., 2005). While some of these factors have been examined in isolation, to date there has yet to be a

comprehensive study that relates these physical characteristics to landing biomechanics in adolescent females. The following sections will highlight the biomechanical differences that emerge with adolescence, and the coinciding sexually dimorphic physical characteristics that also arise at this time.

Biomechanical Differences in Adolescence

As the sex disparity in ACL injury rate begins to emerge during adolescence, more investigators are beginning to examine lower extremity biomechanics in a younger population. These studies have sought to determine if and when sex differences emerge in this population, and the concomitant factors that may potentially drive these changes (Barber-Westin, et al., 2005; Barber-Westin, Noyes, & Galloway, 2006; DiStefano et al., 2011; DiStefano et al., 2010; DiStefano, et al., 2009; Ford, Myer, Brent, & Hewett, 2009; Ford, et al., 2003; Ford, Myer, et al., 2010; Ford, Myer, Toms, & Hewett, 2005; Hass, et al., 2003; Hass, et al., 2005; Hewett, et al., 2004; Hewett, Myer, Ford, & Slauterbeck, 2006; Quatman, et al., 2006; R. J. Schmitz, et al., 2009; Shultz, Nguyen, & Schmitz, 2008; Sigward, Pollard, & Powers, 2008; Sigward, et al., 2011; Swartz, et al., 2005; Yu, et al., 2005). The findings from these studies implicate emerging sex differences in physical characteristics as potential plausible underlying mechanisms for the observed sex differences in biomechanics.

Knee Flexion Angle

The majority of studies that have compared different age and/or maturation groups report a decrease in knee flexion angle at ground contact with age, specifically in females (Hass, et al., 2003; Hass, et al., 2005; Swartz, et al., 2005; Yu, et al., 2005). Hass

and colleagues compared knee biomechanics during landing in pre and post pubescent females using three different drop jump/landing tasks (Hass, et al., 2005). Variations of the task were based on the subsequent movement, if any, from the initial landing (i.e. no additional task after landing, a maximum vertical jump, or a lateral jump). Regardless of the task, knee flexion angle at contact was greater in the pre-pubescent group than the post pubescent group, by an average of 5° (Hass, et al., 2005). Decreased knee flexion at contact for the post pubescent group also held true for jump stride tasks, where instead of falling from a box, participants bound forward from the ground and either "stuck" the landing, performed a vertical jump, or a lateral jump (Hass, et al., 2003). Similarly, a cross sectional study of five, one-year increment age groups performing a two footed stop followed by a maximum jump, showed females landed in a more erect position as age increased from 11 to 16, while males maintained their initial knee flexion angle across all age groups (Yu, et al., 2005).

However, all of these studies reporting decreases in knee flexion angle with maturation are limited to a cross sectional design. In the lone longitudinal study to date that has examined knee flexion angle at ground contact (Ford, Myer, et al., 2010) no sex or maturation differences in knee flexion angle were observed at initial contact when landing from a 31 cm box.

It is possible that in studies that utilized a cross-sectional design to group the subjects by age (Swartz, et al., 2005; Yu, et al., 2005) or maturation (Hass, et al., 2003; Hass, et al., 2005) physical differences were not controlled, and differences between the groups may simply reflect different fitness and strength capabilities of the subject.

Conversely, in the longitudinal study these factors were controlled, as the same subjects were represented throughout and came from the same school district possibly suggesting similar training habits. Additionally, the longitudinal study (Ford, Myer, et al., 2010) utilized a drop jump task while the tasks used in cross sectional studies were often more multi-directional (Hass, et al., 2003; Hass, et al., 2005; Yu, et al., 2005). For these reasons it is difficult to conclude the effect of age on knee flexion angle, as the different tasks are difficult to compare.

Knee Valgus

The growing acceptance of dynamic valgus as a risk factor for ACL injury and rising risk of ACL injury throughout adolescence has led researchers to investigate knee valgus in this population. However, as subjects in this age range are physically changing fairly rapidly, some challenges arise when comparing subjects of different ages. One of the challenges with reporting absolute valgus angle is that the subject's natural alignment is not taken into account. In almost all situations, females will show greater valgus angles, because the average tibiofemoral angle for females is significantly greater than males (Nguyen & Shultz, 2007), and this angle may increase with maturation (Cahuzac, Vardon, & Sales de Gauzy, 1995). Therefore, some report valgus in terms of change of angle, or total movement in the valgus direction, rather than absolute angle. To that end, Ford et al. reported valgus movement of the knee during a drop jump in adolescent male and female athletes as the difference between knee distance at initial contact and maximum valgus (Ford, et al., 2003). Although knee distance at initial contact was not different between the sexes, females displayed more 2D knee valgus motion throughout

ground contact, as evidenced by a 2cm greater change in knee displacement (7.3 cm vs. 5.3 cm) from initial contact to peak knee valgus compared to males (Ford, et al., 2003). Interestingly, females also displayed a higher maximum valgus angle on their dominant side compared to their non-dominant side, though there was no difference in duration of stance phase (Ford, et al., 2003). The authors attributed this to side-to-side imbalances in neuromuscular strength, flexibility, and coordination (Ford, et al., 2003).

To examine changes in knee valgus with maturation, cross sectional designs have been used to compare pre-pubertal, early pubertal, and late/post pubertal males and females (Hewett, et al., 2004; R. J. Schmitz, et al., 2009). Females increased their medial knee displacement from pre and early pubertal to late/post pubertal groups, while the males did not change (Hewett, et al., 2004). This resulted in females having greater knee valgus than males in the late/post pubertal group, but not in the younger groups. This difference can be explained in angular terms by both increased valgus angle at contact (5° vs. 1°), as well as maximum angle (30° vs. 19°) in the most mature female group as compared to the male group (Hewett, et al., 2004). Using a similar task and comparing groups based on Tanner stage of maturation, females were again found to increase the amount of valgus displacement from early to late maturation groups (10° vs. 15.9°), while males decrease the amount of valgus displacement (14.3° to 8.8°) (R. J. Schmitz, et al., 2009). Conversely, when comparing children in a similar age range using videography and a similar task, but not taking maturation into account, others have found no differences between sex or age (Barber-Westin, et al., 2006). The different results may simply reflect the different approaches in grouping participants by age rather than

maturation stage. As males and females of the same age are often at different maturation stages it may be more accurate to stratify subjects by maturation rather than age (Rogol, Clark, & Roemmich, 2000). Collectively these results indicate that females experience greater valgus displacement at the knee during landing as they mature, while males do not (Hewett, et al., 2004; R. J. Schmitz, et al., 2009).

Energetics

As an erect posture at landing has been linked to ACL injury (Koga, et al., 2010; Krosshaug & Bahr, 2005; Krosshaug, Nakamae, et al., 2007), recent attempts have been made to quantify the effect of different lower extremity landing strategies on energy absorption in adults. In a recent study that grouped female participants (11 to 20 years) into high flexion and low flexion groups based on combined hip and knee peak flexion angles the low flexion group absorbed more relative energy at the knee than the hip. Specifically, they used a ratio of knee energy absorption:hip energy absorption, where a value greater than "1" would indicate increased knee energy absorption compared to hip energy absorption while a value of less than "1" would indicate increased hip energy absorption compared to knee energy absorption. The knee to hip energy absorption ratio was 52% greater in the low vs. high flexion landing group (3.5±1.5 vs 2.3±0.8) (Pollard, et al., 2010). The authors suggested that the low flexion group employed a strategy that emphasized the use of the knee extensors over the hip extensors to attenuate impact forces, and that this strategy is less preferable to a more equal distribution of energy absorption about the two joints (Pollard, et al., 2010).

While this way of examining energy absorption further emphasizes the potential detrimental effect of landing in an erect position, how these landing strategies change during adolescence has received little attention. In the only study to date examining energy absorption in maturing groups (Sigward, et al., 2011), females increasingly relied on greater energy absorption of the knee relative to hip during a landing task, particularly from the pre-adolescent to the post-adolescent group (Sigward, et al., 2011). These investigators also examined energy absorption in terms of a ratio of knee energy absorption:hip energy absorption and found that in females the ratio increased from around 2.5 in the pre-pubertal group to just under 3.5 in the post-pubertal group (Sigward, et al., 2011). These results indicate that as females' age, they are using their knee extensors to a greater degree than their hip extensors to decelerate their body, while males use a landing strategy that more evenly distributes the energy absorption across the hip and knee (Sigward, et al., 2011). Moreover, a more flexed position (thought to be protective to the ACL) more evenly distributes the energy absorbed by the knee and hip (Pollard, et al., 2010).

Summary

As males and females physically mature, differences in landing biomechanics begin to arise. The differences that emerge between the sexes mirror those seen in adults with females landing with greater knee valgus (Ford, et al., 2003; Hewett, et al., 2004), a more extended knee (Hass, et al., 2003; Hass, et al., 2005; Yu, et al., 2005), and an increased reliance on the knee extensors to dissipate landing forces (Sigward, et al., 2011). While this change in biomechanics towards higher risk strategies in maturing

females is thought to largely result from changes in physical characteristics that also occur during this time (Ahmad, et al., 2006; Barber-Westin, et al., 2005; Barber-Westin, Noyes, Smith, & Campbell, 2009; Hass, et al., 2003; Quatman, et al., 2006), this relationship has never been directly examined.

Physical Characteristics

Between the ages of 10 and 20, there are many physical changes taking place in the adolescent body and these changes affect young males and females differently. As this is the time when landing biomechanics have also been observed to differentially change in males and females, it is possible that there is a connection between the changing physical characteristics and landing biomechanics. Some of the possible physical factors that may have an impact on biomechanics include body composition, strength, and cardiovascular fitness. While these characteristics have often been implicated when reporting sex differences in biomechanics during maturation, a comprehensive investigation directly examining these factors has yet to be reported. Understanding the influence of these physical characteristics is important, as it may be possible to modify these factors through appropriate training, thereby encouraging more protective movement strategies.

Body Composition

Some of the most drastic physical changes during late childhood and into adolescents are to body composition, with sex differences in body composition starting to materialize around age 12 (Loomba-Albrecht & Styne, 2009). Absolute fat mass is reasonably constant in both males and females prior to puberty, after which females gain

on average 5.6 kg more absolute fat mass than males, at a rate of 1.14 kg of fat mass per year (Loomba-Albrecht & Styne, 2009). Additionally, fat free mass in females begins to plateau around 15-16 years old, while males continue to add fat free mass until 17–19 years of age. The end result is adult females have an average body fat of 28%, whiles adult males have an average of 13% body fat (Heyward & Wagner, 2004).

Body mass index (BMI) is an estimate of body fatness, though it does not directly measure body fat, nor does it give any indication about how body fat is distributed on the body. BMI uses body mass and height (kg/m^2) to categorize individuals as underweight, normal, overweight, or obese and has been utilized by the Centers for Disease Control and Prevention as an overall screening tool for possible health problems. Among injured adolescents, those with a higher BMI have been found to be at greater risk for multiple injuries (Doan, Koehoorn, & Kissoon, 2010). This is consistent with other studies reporting an increased risk of injury in obese children and adolescents compared to their healthy weight counterparts (Bazelmans et al., 2004) and an increased risk for ACL injury in adults with a higher BMI (Uhorchak, et al., 2003). While BMI is an estimate of body fatness, it should be noted that which BMI value constitutes normal, overweight, and obese differ in children and adolescents are compared to adults. Specifically, children and adolescents are in the nutritional status indicator range of "at risk of overweight" when between the 85th and 95th percentile for their age and "overweight" when at or above the 95th percentile (Kuczmarski et al., 2002). An example of the body mass index-for-age percentiles in females age 2 to 20 distributed by the Centers for Disease Control and Prevention can be found in Appendix A.

As body fat has been shown to increase in females throughout adolescence (Heyward & Wagner, 2004; Loomba-Albrecht & Styne, 2009), and increased BMI has been linked to multiple injuries in adolescents (Bazelmans, et al., 2004; Doan, et al., 2010) and ACL injury risk in adults (Uhorchak, et al., 2003), body composition likely plays a role in divergent changes in landing biomechanics and injury risk that occur in adolescent females and males. The role of body composition may be mechanistically related to landing biomechanics by way of weight increasing relative to lean body mass, in turn increasing the demand on the muscles of the lower extremity to control the additional weight. Concomitant with these changes in body composition are changes in muscle strength, which may further compromise functional movement strategies.

Strength

To assess how strength changes throughout maturation and into adulthood, a longitudinal study beginning at age 6 and continuing for 30 years was conducted (Taeymans, Clarys, Abidi, Hebbelinck, & Duquet, 2009). The static strength assessment was measured using an adjustable handgrip dynamometer, which is a portable device that can easily assess strength in any environment. The investigators found strength to be lower in females compared to males at all ages. At age 12, males showed accelerated strength gains while girls demonstrated a deceleration. All groups showed the greatest strength about 1 year after peak height velocity which then declined from age 18 to 35 (Taeymans, et al., 2009). Again, the timing of the divergent strength profiles between the sexes aligns with the timing of divergent landing biomechanics.

While both sexes increase strength as they age, around age 15, relative hamstring strength continues to increase in males but not females (Buchanan & Vardaxis, 2003). This means that when normalizing to total body mass, thus accounting for the mass the muscles of the legs need to control, females are no longer getting stronger in their hamstrings. This plateau of hamstring strength also has implications on the ratio of quadriceps strength to hamstring strength, which has been investigated as a risk factor for ACL injury (Ahmad, et al., 2006; Anderson, Dome, Gautam, Awh, & Rennirt, 2001; Buchanan & Vardaxis, 2003), given their role in controlling anterior tibial displacement at certain knee flexion angles (More, et al., 1993; Yasuda & Sasaki, 1987).

While the lower relative strength in females has been proposed as a possible contributing factor to the females developing at-risk landing biomechanics as they mature (Ahmad, et al., 2006; Barber-Westin, et al., 2005; Barber-Westin, et al., 2009; Hass, et al., 2003; Quatman, et al., 2006), investigators have not identified an association between isokinetic strength of the quadriceps and hamstrings (Barber-Westin, et al., 2006), or a functional strength task of these muscles (R. J. Schmitz, et al., 2009) and landing biomechanics. This may be because sex differences in strength alone cannot fully account for the biomechanical differences between the sexes, suggesting a more comprehensive assessment of the global changes in physical characteristics is needed.

As discussed previously, energy absorption is functionally interpreted as work done on extensor muscles by eccentric muscle action; therefore the decrease in relative quadriceps strength in females may diminish the ability to safely dissipate the energy of landing. This relationship may be particularly important when weight is increasing

relative to lean body mass and strength, thereby increasing the demand on the muscles of the lower extremity. Greater knee extensor strength has been found to predict greater energy absorption at the knee in females (Shultz, et al., 2009), a relationship that is thought to indicate an increased ability of the knee extensors to safely dissipate the energy of landing. It is unknown to this point however, how lower extremity strength is related to energy absorption at the knee relative to other joints of the lower extremity. As the hip and ankle also play an important role in decelerating the body, understanding the relationship between lower extremity strength and energy absorption across all lower extremity joints warrants consideration.

Cardiovascular Fitness

Coinciding with changes in strength and body composition in maturing females is a decrease in cardiovascular fitness compared to males (Janz, et al., 2000; Kemper & Verschuur, 1987; McMurray, et al., 2003; Rutenfranz et al., 1990). Specifically, VO_{2max}, once normalized to body mass, has been shown to decrease 1.7 mL • kg $^{-1}$ • min $^{-1}$ in females from age 14 to 15, representing a decline of 5% of maximum oxygen consumption (McMurray, et al., 2003). In a longitudinal study that assessed cardiovascular fitness each year for 5 years, peak VO_2 decreased 7 mL • kg $^{-1}$ • min $^{-1}$ from pre or early pubertal development to the last testing in females, at which point 75% of the participants were in late or post pubertal development (Janz, et al., 2000). This is a staggering 17.5% decline from 10 to 15 years of age (Janz, et al., 2000). Over the same amount of time, males decreased peak VO_2 by 3 mL • kg $^{-1}$ • min $^{-1}$, representing a decline of only 6% (Janz, et al., 2000). Regardless of age or maturation level, research suggests

that females demonstrate decreased cardiovascular fitness compared to males, even when normalized to body weight (Janz, et al., 2000; Kemper & Verschuur, 1987; McMurray, et al., 2003; Rutenfranz, et al., 1990).

Level of cardiovascular fitness is determined by two main components, genetics (An et al., 2000) and level of training (Hoogeveen, 2000). Individuals with a high level of training are able to work at an increased submaximal VO_2 load for longer periods of time, thus fatiguing at a slower rate than others (Hoogeveen, 2000). In this way, cardiovascular fitness influences resistance to fatigue. As fatigue is related to higher risk landing biomechanics (Borotikar, et al., 2008; Sanna & O'Connor, 2008), fitness has the potential to play a role in injury risk. This contention is based on the fact that injury rates increase near the end of the first and second halves of competition, and is at the highest at the end of games (Price, et al., 2004). Hence, fatigue has been implicated as a potential cause of the increase in injury rates later in practice and competition. To this point, exercise protocols that mimic the intermittent nature of athletics have demonstrated increased knee valgus and rotation during jump landing and cutting in females from pre to post exercise (Borotikar, et al., 2008; Sanna & O'Connor, 2008). If fatigue is the cause of this change in knee joint biomechanics, then it stands to reason that an individual with greater cardiovascular fitness may be more protected from these high risk joint motions than those with less cardiovascular fitness.

While biomechanics have been shown to change following exercise (Borotikar, et al., 2008; Sanna & O'Connor, 2008) and injury rates are highest at the end of practice and competition (Price, et al., 2004), to this point screening for at-risk landing mechanics has

not been done following exercise. As such, assessing the effect of exercise on a drop jump task associated with injury screening (Myer, Ford, Brent, & Hewett, 2012; Myer, Ford, & Hewett, 2011; Myer, Ford, Khoury, Succop, & Hewett, 2010; Padua, et al., 2009; Smith et al., 2012) may have a future clinical impact.

Summary

Late childhood into adolescence is a time of steady growth, with strength and body composition differences between the sexes starting to emerge around 12 years of age. Compared to males, fat free mass and strength in females begins to plateau at around 15-16 years old as they begin to gain significantly more fat mass (Buchanan & Vardaxis, 2003; Loomba-Albrecht & Styne, 2009). These changes suggest that females have less muscle mass and strength to control their body weight and forces imposed on the knee during sport related activity, potentially lending to their increased propensity toward increased knee valgus (Ford, et al., 2003; Hewett, et al., 2004) and greater reliance on knee extensors to absorb a great amount of energy (Sigward, et al., 2011). Coinciding with changes in strength and body composition in adolescent females is a decrease in cardiovascular fitness (Janz, et al., 2000; McMurray, et al., 2003) which may further compromise their dynamic control of knee stability via a lower resistance to fatigue. However, each of these physical characteristics have the potential to be modified somewhat through training. Thus, if these factors are found to effect biomechanics, biomechanics themselves may be improved through training focusing on physical characteristics.

Field Based Measurement Techniques for Assessment of Physical Characteristics

In order to examine associations between physical characteristics and biomechanics, the appropriate measurement tools to accurately assess the physical characteristics of body composition, strength, and cardiovascular fitness are needed. There are many different ways to assess these characteristics, and each measurement tool has advantages and disadvantages. While laboratory based measurements are often considered gold standards, field based assessments have been shown to be highly correlated with many laboratory techniques, and have the added advantage of being portable and more accessible. The latter is particularly beneficial from an injury risk screening and prevention stand point where accurate yet portable and accessible measures are needed to screen large groups of athletes.

The ALPHA (Assessing Levels of Physical Activity) fitness tests battery was developed to provide a valid, reliable, feasible, and safe set of field based tests for the assessment of health-related cardiovascular fitness in children and adolescents (Ruiz, et al., 2011). The field based measures included in the assessment were chosen based on reviews of literature examining the reliability (Artero et al., 2011) and validity (Castro-Pinero, Artero, et al., 2010) of tests that made up existing test batteries published internationally. A total of 7 measures make up the Evidence Based Health-Related Fitness Test Battery (one of three ALPHA options) and include assessments of cardiorespiratory fitness, musculoskeletal fitness (i.e. strength), and body composition (Ruiz, et al., 2011), the modifiable physical characteristics of interest to this study.

Body Composition

Body composition can be directly assessed in a variety of ways including hydrostatic weighing, dual-energy x-ray absorptiometry (DXA), air-displacement plethysmography (i.e. BOD POD), and skinfold thickness. While DXA, BOD POD, and hydrostatic weighing require expensive equipment and must be measured in a laboratory, skinfold thickness assessment can be administered virtually anywhere for a fraction of the cost. Thus, body composition estimates by skin fold assessment are commonly used as a "field" based measure because the calipers are transportable. The methodology is comparatively simple, using specialized calipers to estimate subcutaneous fat thickness. Measurement of skin fold has been validated and found to be accurate when standardized measurement sites and appropriate equipment is used (Lohman, Pollock, Slaughter, Brandon, & Boileau, 1984).

For the ALPHA, the Slaughter equations were chosen as a body fat measurement using skinfold thickness measures over the tricep and subscapular area (Slaughter et al., 1988). The Slaughter equations have been commonly used in studies involving children and adolescents (Freedman et al., 2007; Rodriguez et al., 2005; Steinberger et al., 2005), and reference curves for the measurement sites have been developed (Addo & Himes, 2010). When comparing results of DXA to skinfold thickness, the Slaughter equations for children and adolescents has demonstrated relatively low error and bias (Rodriguez, et al., 2005), are highly correlated to measures obtained by DXA (Steinberger, et al., 2005), and have been recommended for use in clinical settings (Rodriguez, et al., 2005). The low error, high correlation to DXA (considered the gold standard because of the accuracy

of the measurement (Steinberger, et al., 2005)), and ease of clinical application make this method ideal for field testing. As a part of the ALPHA, waist circumference and BMI are also measured.

Strength

The most common form of testing the muscle groups of the lower extremity typically involves using a dynamometer, either handheld (Ahmad, et al., 2006) or isokinetically driven (Buchanan & Vardaxis, 2003). Both methods necessitate specialized equipment, and extensive training of personnel for accurate and reliable measurement. Alternate strategies to test lower extremity strength include jumping or bounding tasks that assess the distance that an individual can explosively propel their body. These tasks need only appropriate space and a ruler type measurement tool to complete. Further, jumping tasks have been found to be predictive of lower extremity strength (Hamilton, Shultz, Schmitz, & Perrin, 2008; Milliken, Faigenbaum, Loud, & Westcott, 2008). Jumping tasks are reliable (Ortega et al., 2008), have strong associations with other tests of muscular strength (Hamilton, et al., 2008), and the standing broad jump has been suggested as a general index of muscular fitness in a youth population (Castro-Pinero, Ortega, et al., 2010). As such, the ALPHA uses the standing broad jump to assess lower extremity strength.

The ALPHA also assesses upper extremity strength using handgrip. Upper extremity strength evaluation is included in the ALPHA as handgrip has been shown to be a reliable assessment of overall strength in children (Milliken, et al., 2008). While hand grip strength is included in the ALPHA, and is considered a reliable assessment of

strength in children (Milliken, et al., 2008), the drop jump task is a task with demands exclusive to the lower extremity. Given the lower extremity specific demands of the drop jump task, the high correlation of grip strength with broad jump distance (Milliken, et al., 2008), and its lower equipment burden, the standing broad jump is the measure of choice to identify a relationship between strength and landing biomechanics.

Cardiovascular Fitness

While VO_{2max} testing is considered the gold standard to assess cardiovascular fitness, this method, as with other laboratory measures, is both time consuming and requires specialized equipment. However, VO_2 estimate via performance on a progressive 20 meter shuttle run test has been highly correlated to directly measured VO_{2max} (Leger, Mercier, Gadoury, & Lambert, 1988; Ramsbottom, Brewer, & Williams, 1988). This allows cardiovascular fitness to be easily assessed as a part of multi-stage field based measurement tool, and it is this test that is used in the ALPHA.

The 20 meter shuttle run was chosen for the ALPHA after the investigators assessed the criterion validity of the 20 meter shuttle run (Ruiz, et al., 2008; Ruiz et al., 2009), the one mile run/walk (Castro-Pinero, Mora, Gonzalez-Montesinos, Sjostrom, & Ruiz, 2009) and the ½ mile run/walk tests (Castro-Pinero, Ortega, Mora, Sjostrom, & Ruiz, 2009). Using the performance on the 20 meter shuttle run and the subjects' sex, age, weight, and height, the investigators developed an artificial neural network (ANN) equation as an estimation of maximal oxygen uptake (VO_{2max}) (Ruiz, et al., 2008). This equation was validated against directly measured oxygen uptake in an adolescent population, and compared against Leger's equation (Leger, et al., 1988). The ANN

equation was found to be the most accurate of the current prediction equations available, and is suggested for use in the adolescent population (Ruiz, et al., 2008).

Summary

There are many different ways to assess body composition, strength, and cardiovascular fitness. While laboratory based measurements are often considered gold standards, field based assessments have shown to be highly correlated to the laboratory assessments. Advantages of field based assessments of these modifiable physical characteristics over laboratory assessments include lower cost of equipment, less equipment, time efficiency, and ease of administration to multiple subjects in a single session. To this end, the ALPHA has been shown to be a valid, reliable, feasible, and safe set of field based tests for the assessment of health-related cardiovascular fitness in children and adolescents (Ruiz, et al., 2011). The seven measures that make up the Evidence Based Health-Related Fitness Test Battery include assessments of cardiorespiratory fitness (20 meter shuttle run), musculoskeletal fitness (standing broad jump and handgrip), and body composition (BMI and skinfold) (Ruiz, et al., 2011). In addition to requiring less equipment, all of the field tests included in the ALPHA can be completed in any space large enough to the set up the fitness components (i.e. the 20 meter shuttle run), and can therefore be administered in any gymnasium.

The field based tests included in the ALPHA are promising in ACL injury research because of the potential to screen large numbers of adolescent athletes. As such, the 20 meter shuttle run, standing broad jump, and skinfold assessment are ideal for use in identifying relationships between modifiable physical characteristics and high risk

landing biomechanics. If associations between modifiable physical characteristics and high risk landing biomechanics are identified, both of which have been shown to increase throughout adolescence, these field based assessments of physical characteristics have great potential for use in screening of adolescent athletes in the future. Appropriate identification of individuals with high risk landing biomechanics will lead to early, suitable, and targeted intervention strategies in what may be a group at a higher risk for injury.

Summary

The risk of anterior cruciate ligament (ACL) injury increases steadily in females after age 12 (Gianotti, et al., 2009; Le Gall, et al., 2006; Peterson, et al., 2000), and peaks around age 15 (Csintalan, et al., 2008; Shea, et al., 2004). As a result, adolescence and adult females suffer ACL injuries at a rate of 4-6 times greater than males (E. Arendt & Dick, 1995; E. A. Arendt, et al., 1999; Deitch, et al., 2006; Gomez, et al., 1996) . During the time when sex differences in ACL injury rates begin to emerge, knee biomechanics in females trend toward higher risk strategies, including increased knee valgus (Ford, et al., 2003; Hewett, et al., 2004), a more extended knee (Hass, et al., 2003; Hass, et al., 2005; Yu, et al., 2005), and increased reliance on the knee extensor muscles to dissipate landing forces (Sigward, et al., 2011). This propensity for developing higher risk knee joint biomechanics is thought to largely result from changes in physical characteristics that also occur during this time (Ahmad, et al., 2006; Barber-Westin, et al., 2005; Barber-Westin, et al., 2009; Hass, et al., 2003; Quatman, et al., 2006).

Late childhood and adolescence is a time of steady growth, with strength (Buchanan & Vardaxis, 2003) and body composition (Loomba-Albrecht & Styne, 2009) differences between the sexes starting to materialize around age 12 followed shortly by decreased fitness levels (Janz, et al., 2000; McMurray, et al., 2003). Specifically, maturing females develop greater fat mass (Loomba-Albrecht & Styne, 2009) as strength gains plateau (Buchanan & Vardaxis, 2003) and cardiovascular fitness decreases (Janz, et al., 2000) while maturing males increase lean mass and strength while maintaining fitness (Buchanan & Vardaxis, 2003; Janz, et al., 2000; Loomba-Albrecht & Styne, 2009). Given these concomitant changes, and the theories put forth regarding changing mechanics resulting from changes in physical characteristics (Ahmad, et al., 2006; Barber-Westin, et al., 2005; Barber-Westin, et al., 2009; Hass, et al., 2003; Hewett, et al., 2004; Quatman, et al., 2006; Wild, Steele, & Munro, 2012; Yu, et al., 2005), there is a need to determine if changes in physical characteristics during maturation are associated with the changes in knee biomechanics observed during maturation. Moreover, it is important to assess these physical changes via field tests that can be effectively used to screen for those at future risk for injury. Before progressing on to costly and time intensive longitudinal studies that track individual changes in physical characteristics and landing biomechanics over time, it is reasonable first step to examine how modifiable characteristics are associated with landing biomechanics in a representative group of adolescent females. Understanding these associations may improve our understanding of the underlying factors that promote high risk landing strategies in maturing athletes, and

thereby what we should target in our injury risk screening and intervention strategies for this age group.

CHAPTER III

METHODS

The objective of this research was to examine the extent to which modifiable factors that change with maturation effect knee joint biomechanics during a landing task. The approach was to measure cardiovascular fitness, strength, and body composition in physically active adolescent females (ages 11-15), and examine the extent to which these factors predict high risk knee biomechanics, before and after an exercise challenge. Females were chosen to control for hormonal (Oerter, et al., 1990; Rose, et al., 1991) and anatomical (Cahuzac, et al., 1995) differences that occur between the sexes during this time of rapid growth and development. The central hypothesis was that higher body fat percentage, lower strength, and lower fitness would predict greater knee valgus, greater tibial rotation, and greater energy absorption at the knee relative to the hip and ankle during landing.

Participants

Fifty (50) healthy, physically active, adolescent females who are equally distributed between the ages of 11 and 15 were recruited from local sports teams and clubs as well as middle school athletic programs. Only females were chosen because they represent a group with wide variability in their physical characteristics (Buchanan & Vardaxis, 2009; Janz, et al., 2000; Kuczmarski, et al., 2002). Physically active was defined as an individual currently active in sport activities that include running and

cutting and landing maneuvers such as soccer, basketball, and lacrosse. Inclusion criteria were: 1) current engagement in sport activities at least 3 days per week; and 2) no current lower extremity injury. Exclusion criteria were: 1) previous history of injury to the capsule, ligaments, or menisci of either knee; 2) vestibular or balance disorders that could cause them to lose their balance during functional tasks; and 3) cardiovascular disease. All guardians read and signed a consent form, and all participants read and signed an assent form. Both forms were approved by UNCG's Institutional Review Board for the Protection of Human Subjects (Appendix B). Each participant attended a single testing session consisting of the field physical assessment measurements and a biomechanical analysis of the lower extremity during a drop jump task before and after an exercise challenge. Participants were instructed to avoid strenuous activity prior to their test day that was beyond what they normally and consistently performed. This was intended to limit muscle soreness or other changes in muscle tension which may confound the study findings. For biomechanical analysis, participants were outfitted in a standardized athletic shoe and custom compression shorts to allow attachment of LED markers to the thigh.

Procedures

Upon arrival, written consent and assent was obtained, and subject demographics of age, sex, race, ethnicity, height, mass, hip circumference, and waist circumference were recorded. Participants also completed physical activity (type, duration, and intensity), injury history (Appendix C), and self-reported Tanner stage of maturation questionnaires (Morris & Udry, 1980)(Appendix D). Participants were then familiarized

to the drop jump, underwent skinfold measurements, and were instrumented and digitized for motion capture data collection. Once instrumented, they completed a dynamic warm-up, standing broad jump, and biomechanical assessment of a drop jump before and after a fitness test, which served as the exercise challenge. The data collection procedures for each of the field tests for modifiable physical characteristics and lower extremity biomechanical assessment follow.

Demographics and Questionnaires

Subject demographics of age, sex, race, ethnicity, height, mass, hip and waist circumference were recorded, and physical activity (type, duration, and intensity) and injury history were assessed with a standard questionnaire and entered into Excel. Mass was measured with a standard laboratory scale (Seca, Hamburg, Germany) in kilograms while height, hip circumference, and waist circumference were measured with a tape measure in centimeters.

To assess Tanner stage of maturation, participants were given a standardized series of drawings with explanatory text to assess their own pubertal development (Morris & Udry, 1980). The drawings consisted of the 5 stages of breast development and 5 stages of female pubic hair development with appropriate written descriptions accompanying the drawings. Participants were asked to select the drawing and stage that best indicated her own development. A single individual score, ranging from 1 (pre pubertal) to 5 (adult level of development), was computed by averaging the two ratings (Rapkin, Tsao, Turk, Anderson, & Zeltzer, 2006). To minimize any discomfort associated with completing the self-assessment, the questionnaire was completed in a

private office. The questionnaire was given to the participants in a plain manila envelope and they were instructed to place the questionnaire back in the envelope and leave it in the office upon completion. The less intrusive self-reported Tanner stage assessment has been previously validated in children and adolescents (Leone & Comtois, 2007; K. E. Schmitz, et al., 2004), and has been used previously by the Applied Neuromechanics Research Laboratory (R. J. Schmitz, et al., 2009; Shultz, et al., 2008).

Familiarization of the drop jump task

Familiarization of the drop jump task first consisted of the investigator demonstrating and giving verbal instructions regarding the successful completion of the task. The participant was then asked to perform the task by standing on a 45 cm box, with feet shoulder width apart, hips and knees extended, toes facing forward, equal weight on both feet and hands at ear level. The participant was considered comfortable with the task after completing 5 consecutive successful trials, though additional practice was given if requested. A trial was deemed successful if the participant: 1) slid off the box without jumping or stepping; 2) landed with one foot on each force plate both prior to and following the maximal jump; 3) produced a maximal effort; and 4) kept their hands at ear level.

Field Tests of Modifiable Physical Characteristics

The ALPHA (Assessing Levels of Physical Activity) health-related fitness test battery was developed to provide a set of field based tests for the assessment of health-related cardiovascular fitness in children and adolescents that was valid, reliable, feasible, and safe (Ruiz, et al., 2011). The field based measures were chosen based on minimal

instrumentation requirements and reviews of literature examining the reliability (Artero, et al., 2011) and validity (Castro-Pinero, Artero, et al., 2010) of tests that made up existing test batteries. The ALPHA consists of a 20 meter shuttle run, standing broad jump (SBJ), handgrip, and skinfold assessments. However, handgrip strength was not included in the current study. Given the lower extremity specific demands of the drop jump task and high correlation of grip strength with broad jump distance (Milliken, et al., 2008), SBJ was deemed the most appropriate strength measure to include in this study.

Body composition (%BF) was assessed via measurements of the triceps and subscapular skinfolds thickness using a research grade caliper (Harpenden; West Sussex, UK) in accordance with the Slaughter equations (Ruiz, et al., 2011; Slaughter, et al., 1988). When comparing results of skinfold measures, Slaughter (Slaughter, et al., 1988) equations for children and adolescents have shown high correlations with DXA (Treuth, Butte, Wong, & Ellis, 2001) which has been validated against CT (Kaul et al., 2012) and MRI measures (Bridge et al., 2011). These equations have also demonstrated relatively low error and bias, and have been recommended for use in clinical settings (Rodriguez, et al., 2005). Three measurements were taken at each site and recorded in mm. The average of the three values were entered into Excel and input into the Slaughter equation for calculation of %body fat (Slaughter, et al., 1988):

For skinfolds of <35 mm = 1.33(sum of 2 skinfolds)-0.013(sum of 2 skinfolds)^2-2.5

For skinfolds of >35 mm = 0.546(sum of 2 skinfolds)+9.7

Prior to the study, the principal investigator established high between day reliability when measuring these sites in 11 adolescent females (Tricep ICC: 0.98 SEM: 0.78 mm Subscapular ICC: 0.99, SEM: 0.38 mm)(Table 1).

Table 1. Between Day test-retest Reliability analysis of Field Tests
[Note: SEM is expressed in units of the measure. Thus, Triceps and Subscapular skinfolds are expressed in millimeters (mm), and Broad Jump in centimeters (cm).]

Variable	Day 1 Mean ± SD	Day 2 Mean ± SD	ICC (2,k)	SEM
Triceps	18.5 ± 7.4	18.8 ± 6.9	0.98	0.78
Subscapular	16.6 ± 6.2	16.8 ± 5.9	0.99	0.38
Broad Jump	157.2 ± 18.0	158.3 ± 14.1	0.97	2.94

Lower body strength was assessed via standing broad jump (SBJ). Prior to SBJ, participants completed a standardized dynamic warm up that has been used in previous research (Cone et al., 2011). The dynamic warm up consisted of 3-minutes of jogging and low-intensity running followed by approximately 9-minutes of dynamic flexibility movements of increasing complexity and running at increasing intensities (Table 2). This lower extremity warm up systematically and progressively targeted muscle groups from distal to proximal.

Table 2. Dynamic Warm-Up

ACTIVE WARMING PHASE (3 minutes)
Jogging
Backwards Jog
Side Shuffle
DYNAMIC FLEXIBILITY PHASE (7-9 minutes)
Heel-Toe Walks (forwards)
Walking Calves-Straight Leg (forwards)
Easy-Alternate Leg Heel Kicks (forwards)
Progress-Alternate Leg Heel Kicks (forwards)
Single-Leg Heel Kicks (forwards)
Walking Hamstrings (forwards)
Side Shuffle (alternating-forwards)
Side Shuffle (alternating-backwards)
Walking Lateral Lunge (continuous)
Open the Gate (forwards)
Close the Gate (forwards)
High Knees (forwards)
High Knees (backwards)

Jump distance has been shown to be a valid predictor of isokinetic strength (Hamilton, et al., 2008) and has been recommended for assessment of muscular strength in children and adolescents (Castro-Pinero, Ortega, et al., 2010). Participants began standing on both feet at a start position and were instructed to jump forward as far as possible. Distance from the start position to the back of the heel closest to the start position was measured in centimeters (cm). Participants were given three practice trials, unless jump distance increased with each trial. In this case, subsequent practice trials were allowed until jump distance stabilized. Maximal effort was confirmed by stabilization of jump distance along with low variability between the final three trails. The average distance of three jumps were recorded (Castro-Pinero, Ortega, et al., 2010).

Cardiovascular fitness was assessed with a 20 meter shuttle run test. The 20 meter shuttle run has been validated in participants of different ages and both sexes (Liu, Plowman, & Looney, 1992; Ruiz, et al., 2009), and the VO_2 estimate is reported to be highly correlated with direct laboratory measurement of VO_{2max} (Leger, et al., 1988; Ramsbottom, et al., 1988; Ruiz, et al., 2008). The 20 meter shuttle run consisted of running between two cones spaced 20 meters apart at a pace dictated by pre-recorded audio cues. The test continued until the participant failed to keep pace on two consecutive shuttle runs. The last stage completed was scored and recorded as total distance in meters, as was the stage completed to the nearest half stage (see Table 3).

Table 3. Speeds, stages, and distances of 20 meter shuttle run

Speed (km/h)	**Stage**	**Distance**									
8.5	**1**	20	40	60	80	100	120	140			
9	**2**	160	180	200	220	240	260	280	300		
9.5	**3**	320	340	360	380	400	420	440	460		
10	**4**	480	500	520	540	560	580	600	620		
10.5	**5**	640	660	680	700	720	740	760	780	800	
11	**6**	820	840	860	880	900	920	940	960	980	
11.5	**7**	1000	1020	1040	1060	1080	1100	1120	1140	1160	1180
12	**8**	1200	1220	1240	1260	1280	1300	1320	1340	1360	1380
12.5	**9**	1400	1420	1440	1460	1480	1500	1520	1540	1560	1580

The artificial neural network (ANN) equation was then be used to estimate VO_{2max}. The ANN was developed as an estimation of VO_{2max} (Ruiz, et al., 2008) that was validated against directly measured oxygen uptake in adolescent population, and has been found to have the lowest error of available prediction equations (Ruiz, et al., 2009). The equation takes the following variables into account: sex, age, weight, height, and distance run. The variables were entered into an Excel template for computation of VO_{2max} (ml/(kg min)) (Ruiz, et al., 2008):

VO_{2max} (ml/(kg min)) =

(1/(1+exp(-(1/(1+exp(-((A1*0.8+(-0.7))*(-1.03329) +
(B1*0.114285714286+(-1.38571428571))*0.54719 +
(C1*0.012213740458+ (-0.406870229008)) * 0.61542 +
(D1*0.0195598978221+(-2.76356892177))*(-0.51381) +
(E1*0.0842105263158+(-0.0684210526316))*(-0.92239) + (-0.34242))))
*(-0.95905)+1/(1+exp(-((A1*0.8+(-0.7))*(-1.19367) +
(B1*0.114285714286+(-1.38571428571))*(-1.54924) +
(C1*0.012213740458+(-0.406870229008))*(-3.18931)+
(D1*0.0195598978221+(-2.76356892177))*0.77773+
(E1*0.0842105263158+(-0.0684210526316))*3.31887+ (-0.55696)))) *
2.19501+1/(1+exp(-((A1*0.8+(-0.7))*1.38191+(B1*0.114285714286 + (-
1.38571428571)) * (-2.14449)+(C1*0.012213740458 + (-
0.406870229008)) *0.0485+(D1*0.0195598978221+(-2.76356892177))
*0.10879+(E1 *0.0842105263158+(-0.0684210526316)) *(-
4.90052)+0.53905))) *(-2.567)+(-0.05105))))-(-0.478945173945))
/0.0204587840012

A1 = sex (boys = 1; girls = 2); B1 = age (year, age range 12 - 19 years);
C1 = weight (kg); D1 = height (cm); E1 = stage (0.5)

Because of the fatiguing nature of this fitness test, the 20 meter shuttle run also served as the exercise challenge, before and after which lower extremity

biomechanics were assessed (see next section). Time between completion of the shuttle run and drop jumps was less than 60 seconds.

Lower Extremity Biomechanical Assessment

Using methods previously described (R. J. Schmitz & Shultz, 2010; Shultz, et al., 2009), participants performed 3 maximal drop jumps while fully instrumented for biomechanical assessment, prior to and following the shuttle run. Participants were instrumented with optical LED marker clusters (Phase Space, San Leandro, CA), placed on the left foot, shank, thigh, and sacrum. Sacrum markers were attached using double sided tape. The thigh and shank segment markers were attached using hook and loop material on specially designed compression shorts and a shank sleeve, while the foot segment markers were taped to the left shoe. Joint centers were determined as the midpoint of the medial and lateral malleoli (ankle), midpoint of the medial and lateral femoral epicondyles (knee), and using the Bell method (hip) (Bell, Brand, & Pedersen, 1989). The un-tethered, active marker system allowed the freedom to fully instrument, digitize, and collect data on a subject prior to physical activity, conduct the fitness test, and return to the capture space to collect additional data without having to re-digitize the participant or re-calibrate the capture space.

To perform the drop jump, participants began the task in a standardized take-off position with the front edge of their shoes aligned along the front edge of the platform and their hands placed at ear level. As instructed in the familiarization, participants were asked to drop off the platform, without jumping or stepping, and perform a maximal vertical jump upon landing. No special instructions were given regarding drop jump

biomechanics to prevent experimenter bias. The participant was instructed to keep their hands at ear level to eliminate variability due to arm-swing and to limit marker obstruction. Trials were deemed unsuccessful and repeated if the participants lost their balance, their hands dropped below the level of the ears, or they did not land back on the force plate following the maximum vertical jump. The drop jump task has been previously used to assess landing biomechanics in this population (Ford, et al., 2003; Ford, Myer, et al., 2010; Ford, Shapiro, et al., 2010; R. J. Schmitz, et al., 2009).

Biomechanical data was processed using MotionMonitor Software (Innovative Sports Training, Chicago, IL USA) and custom MATLAB (MathWorks, Natick, MA USA) code. Kinematic (240 Hz) data were linearly interpolated to kinetic (1000 Hz) data and exported unfiltered. Joint angle and moment data were low-pass filtered at 12 Hz using a 4th order, zero lag Butterworth filter, while ground reaction force data were low-pass filtered at 60 Hz using a 4th order, zero lag Butterworth filter within the MATLAB code. A segmental reference system quantified the 3-D kinematics, while Euler's equations described joint motion about the defined axes in the pelvis, thigh, and shank with a rotational sequence of Z (flexion/extension) Y' (internal/external rotation) X" (abduction/adduction). The following motions were positive regardless of the joint; flexion, internal rotation, and adduction. Data were collected for 3 seconds (0.5 seconds prior to ground contact (>10N vGRF) and 2.5 seconds following ground contact). Kinetic and kinematic data of the hip, knee, and ankle were extracted from initial ground contact (defined as >10N vGRF) to peak knee flexion. Intersegmental kinetic data were calculated via an inverse dynamic approach and resultant internal moments were

normalized to weight and height (Nm x-BW^{-1}xHt^{-1}). Net joint powers were calculated as the product of the normalized joint moment and joint angular velocity at each time point. Work done on the extensor muscles was then calculated by integrating the negative portion of the joint power curve, and reported as normalized to body weight and height (Joules x BW^{-1} x Ht^{-1}) consistent with previous work examining energetics during landing (Montgomery, et al., 2012; Shultz, Schmitz, Tritsch, & Montgomery, 2012).

From these data the following variables were extracted and used for data analysis: knee valgus angles, tibial rotation angles, and energy absorption at the hip, knee, and ankle. Frontal and transverse plane knee angles were recorded for initial ground contact, peak displacement, and total excursion (peak – initial). Relative energy absorption at the knee, calculated as the percentage of knee work to total work (hip work + ankle work + knee work) was also recorded. The average of three trials was calculated for each variable, for both the pre-exercise and post-exercise test conditions.

Statistical Plan

Age, height (cm), mass (kg), maturation stage, percent body fat (%BF), standing broad jump distance (cm), estimated VO_{2max} (ml/(kg min)), and the biomechanical variables of dynamic knee valgus and tibial rotation (initial, peak, and excursion), and relative energy absorption at the knee, were entered into Excel and transferred to SPSS for later analysis.

Hypothesis 1: One or a combination of greater body fat, decreased strength, and decreased cardiovascular fitness, will predict greater knee valgus, greater tibial rotation, and greater relative energy absorption at the knee, during the deceleration phase of

landing. To test hypothesis 1, a series of multiple linear regression analyses were used. A separate analysis was performed for each dependent variable [knee valgus (initial, peak, and excursion), tibial rotation (initial, peak, and excursion), relative energy absorption at the knee]. In each model, maturation stage was first entered and controlled for, and then the same three of modifiable physical characteristics (%BF, SBJ (cm), VO_{2max}) were simultaneously entered into the model on the second step. Thus, a total of 4 variables were entered into each model.

Hypothesis 2: Following an exercise challenge, one or a combination of greater body fat, decreased strength, and decreased cardiovascular fitness, will predict greater knee valgus, greater tibial rotation, and greater relative energy absorption at the knee, during the deceleration phase of landing. To test hypothesis 2, the same series of multiple linear regression analyses was used as for Hypothesis 1, with the exception that the dependent variables were obtained after the exercise challenge. A separate analysis was performed for each dependent variable [knee valgus (initial, peak, and excursion), tibial rotation (initial, peak, and excursion), relative energy absorption at the knee]. In each model, maturation stage was first entered and controlled for, and then the same three modifiable physical characteristics (%BF, SBJ (cm), VO_{2max}) were simultaneously entered into the model on the second step. Thus, a total of 4 variables were entered into the model.

Hypothesis 3: The relationship between the predictor variables (one or a combination of greater body fat, decreased strength, and decreased cardiovascular fitness) and knee joint biomechanics will be stronger when knee joint biomechanics are

acquired after the exercise challenge compared to when acquired before the exercise challenge. To test hypothesis 3, the regression equations testing hypothesis 1 and hypothesis 2 were compared. Descriptive differences regarding variance explained and standardized coefficient weights were noted. To allow direct comparisons between all models pre and post exercise, all physical characteristics (body fat, strength, and cardiovascular fitness) were entered simultaneously with no stepwise eliminations performed when testing hypotheses 1 and 2. Additionally, a multi-level model exploratory analysis was used to statistically test the difference in those coefficients. The multi-level model controlled for repeated measurements on each person and allowed for the addition of interaction terms with time (pre and post exercise). Specifically, the interaction terms of %BF x exercise, SBJ x exercise, and VO_{2max} x exercise were added to the model. This was accomplished by coding exercise as a dummy variable, where "0" indicates pre exercise and "1" indicates post exercise. The resulting model was:

$$\textbf{Dep. Variable} = \beta_0 + \beta_1\textbf{Tanner stage} + \beta_2\%\textbf{BF} + \beta_3\textbf{SBJ} + \beta_4 + \beta_5\textbf{Exercise} + \beta_6\textbf{Tanner stage*Exercise} + \beta_7\ \%\textbf{BF*Exercise} + \beta_8\textbf{SBJ*Exercise} + \beta_9\textbf{VO}_{2max}\textbf{*Exercise}$$

It was acknowledged that the constraints of the current study would likely render the multi-level model underpowered to detect interaction effects. However, while exploratory in nature, this model was included as a direct test of the effect of exercise.

All hypotheses were evaluated at $P \leq .05$. Power analysis revealed that a sample size of 46 would achieve 80% power to detect an R-Square value of 0.20 attributed to the

3 independent variables of interest (%BF, SBJ (cm), VO_{2max}) using an F-Test. The variables tested were adjusted for an additional independent variable (maturation stage) with an R-Squared of 0.05 (Cohen, 1988). To ensure adequate power, a sample size of 50 participants was used. An R-Squared of 0.20 was chosen based on similar data from a previous project on adults using the dependent variable of relative energy absorption at the knee and predictors of quadriceps strength, percent body fat via DXA, and fitness via distance run on the Yo-Yo Intermittent Recovery Test Level 1 (Shultz, et al., unpublished data). These data were chosen as the closest surrogates to the field-based measures of the current study.

CHAPTER IV

RESULTS

Fifty-one adolescent female subjects participated and completed the testing. Data from one participant was not included as her body fat percentage (Tanner stage 2 with a Body Fat 42.9%; 3 SD above the mean) was determined to be an extreme outlier. Though she met all inclusion criteria, her BMI percentile for her age was greater than the 95th percentile. This was well above the average of the 53rd percentile for all other participants, as well as the 45th percentile for the participants that were ± 1 Tanner stages from her. Exclusion of her data did not change any of the relationships between physical measures of strength, body composition, and cardiovascular fitness and the dependent variables examined for hypotheses 1 and 2, though her data did influence the relationships between the physical measures. Therefore, data from 50 participants (Age: 12.7±1.4yrs, Tanner stage: 3.4±0.8, Height: 160.7±7.8 cm, Mass: 52.3±10.2 kg) were used for analyses. Overall Mean, SD and range for physical measures are listed in Table 4 and physical measures by Tanner stage in Table 5. Values obtained for the dependent variables before and after exercise are listed in Table 6 along with p-values noting significance from the repeated measures ANOVA. A correlation matrix of the independent and dependent variables before and after exercise is presented in Table 7. Histograms of dependent variables are included in Appendix E (pre-exercise) and

Appendix F (post-exercise), and histograms of physical measures are in Appendix G. SPSS outputs of all models used are in Appendix H.

Table 4. Descriptive Statistics for Physical Measures

Measure	Mean (SD)	Range
Tanner Stage	3.41 (0.82)	1.5 - 4.5
Broad Jump *(centimeters)*	151.49 (17.38)	114.67 – 193.34
VO_{2max} *(ml/(kg min))*	33.46 (5.78)	27.78 - 56.26
Body Fat *(%)*	22.82 (6.07)	10.70 – 38.46
Activity Rating	14.10 (2.50)	8 - 16

Table 5. Descriptive data of physical measures by Tanner stage (Mean (SD))

Tanner Stage	N	SBJ cm	VO^2 max ml/(kg min)	Body Fat %	Activity Rating
1.5	2	159.2 (12.0)	36.9 (1.1)	17.0 (8.9)	16.0 (0.0)
2.0	2	143.2 (17.7)	30.2 (0.2)	21.4 (4.1)	12.5 (0.7)
2.5	8	148.5 (19.1)	32.9 (5.5)	22.5 (6.6)	13.5 (2.6)
3.0	8	150.7 (14.6)	36.9 (9.1)	18.9 (5.0)	14.0 (2.5)
3.5	9	150.8 (17.0)	31.7 (3.0)	22.1 (6.9)	13.8 (2.8)
4.0	13	148.1 (17.2)	32.9 (5.4)	25.7 (5.6)	14.6 (2.2)
4.5	8	161.7 (21.1)	33.5 (5.9)	24.9 (4.4)	14.3 (3.3)

Table 6. Descriptive Statistics for Dependent Variables before and after Exercise

Measure	Before Exercise Mean ± SD (Range)	After Exercise Mean ± SD (Range)	P value
Initial Knee Valgus *(degrees)*	1.06 ± 4.15 (-10.24 - 8.57)	1.27 ± 4.03 (-8.95 – 8.55)	.531
Peak Knee Valgus *(degrees)*	-9.18 ± 7.06 (-24.77 – 7.05)	-9.15 ± 7.22 (-27.97 – 5.35)	.968
Knee Valgus Excursion *(degrees)*	-10.24 ± 5.53 (-21.99 – 0.00)	-10.42 ± 5.73 (-20.91 – -0.63)	.758
Initial Tibial Rotation *(degrees)*	-3.80 ± 8.66 (-24.96 – 19.25)	-5.17 ± 8.97 (-26.82 – 16.20)	.014
Peak Internal Tibial Rotation *(degrees)*	8.26 ± 8.09 (-15.39 – 22.94)	8.57 ± 8.92 (-17.24 – 25.11)	.594
Internal Tibial Rotation Excursion *(degrees)*	12.06 ±6.72 (0.025 -26.83)	13.74 ± 7.12 (0.00 – 29.56)	.019
Relative Knee Absorption *(%)*	0.34 ± 0.15 (0.01 – 0.67)	0.32 ± 0.15 (0.05 – 0.64)	.159

Table 7. Correlation Matrix of Dependent and Independent Variables with above diagonal values after exercise and below diagonal before exercise. Values on the diagonal are pre and post exercise correlations.

			After Exercise										
		Tanner stage	**Broad Jump**	**VO_2**	**Body Fat**	**Activity Rating**	**In. Valgus**	**Pk Valgus**	**Valgus Ex**	**In. Rotation**	**Pk Rotation**	**Rotation Ex**	**Relative EA**
Before Exercise	**Tanner**	**1**	0.115	-0.071	0.331*	0.066	0.287*	-0.001	-0.203	0.067	0.092	0.065	0.133
	Broad Jump	0.115	**1**	0.280*	-0.177	0.081	0.150	0.223	0.175	0.280*	0.351*	0.119	0.027
	VO_2	-0.071	0.280*	**1**	-0.556*	0.214	0.011	0.150	0.181	0.114	0.094	-0.039	0.009
	Body Fat	0.331*	-0.177	-0.556*	**1**	0.042	-0.031	-0.250	-0.294*	-0.181	-0.184	0.021	-0.002
	Activity Rating	0.069	0.082	0.213	0.041	**1**	-0.021	0.056	0.085	-0.133	-0.064	0.109	0.151
	In. Valgus	0.197	0.119	0.045	-0.080	0.099	**0.832***	0.611*	0.068	-0.463*	-0.098	0.466*	0.053
	Pk Valgus	-0.055	0.224	0.068	-0.262	0.112	0.623*	**0.757***	0.831*	-0.306*	0.040	0.401*	-0.056
	Valgus Ex	-0.219	0.197	0.053	-0.275	0.068	0.045	0.809*	**0.726***	-0.060	0.119	0.178	-0.108
	In. Rotation	0.012	0.208	0.101	-0.125	-0.074	-0.510*	-0.240	0.076	**0.908***	0.705*	-0.348*	0.085
	Pk Rotation	0.129	0.428*	0.155	-0.156	-0.057	-0.087	0.080	0.167	0.709*	**0.894***	0.409*	0.071
	Rotation Ex	0.179	0.282*	0.041	-0.004	0.049	0.563*	0.426*	0.121	-0.407*	0.341*	**0.766***	0.015
	Relative EA	-0.051	0.215	0.152	-0.152	0.053	0.039	-0.044	-0.086	0.181	0.242	0.071	**0.828***

* Indicates significant at $p<0.05$

Hypothesis 1: Contribution of Physical Measures to Landing Biomechanics prior to an Exercise Challenge

The following results describe the relationship between physical measures of strength, body composition, and fitness once accounting for Tanner stage on lower extremity biomechanics when landing biomechanics were measured under resting conditions (i.e. prior to the exercise challenge).

Knee Valgus

When predicting knee valgus prior to exercise, Tanner stage accounted for 3.9% (p=0.15), 0.3% (p=0.70), and 4.8% (p=0.13) of variance in initial valgus, peak valgus, and valgus excursion respectively. Once accounting for Tanner stage, the addition of the physical measures of strength, body composition, and cardiovascular fitness did not significantly increase the predicted variance in initial valgus (R^2 change=3.0%, p=0.70), peak valgus (R^2 change=11.7%, p=0.13), or valgus excursion (R^2 change=9.8%, p=0.18). The final models can be found in Table 8.

Tibial Rotation

When predicting internal tibial rotation prior to exercise, Tanner stage accounted for 0.0% (p=0.94), 1.7% (p=0.37), and 3.2% (p=0.21) of variance in initial internal tibial rotation, peak internal tibial rotation, and internal tibial rotation excursion, respectively. Once accounting for Tanner stage, the addition of the physical measures of strength, body composition, and cardiovascular fitness increased the predicted variance in peak tibial rotation (R^2 change = 18.7%, p=.02) but not in initial tibial rotation (R^2 change = 5.2%, p=0.49) and rotation excursion (R^2 change = 7.1%, p=0.33). Of the variables included, standing broad jump distance was the sole significant predictor for peak internal tibial

rotation (Table 8). The unstandardized regression coefficient (β=0.18) indicates that for every 1 cm greater in broad jump distance, peak internal tibial rotation was 0.18 degrees greater. In more clinically relevant terms, for every 10 cm greater distance jumped, a participant had a predicted 1.8° greater peak internal rotation.

Relative Knee Energy Absorption

When predicting relative knee energy absorption, Tanner stage accounted for 0.3% variance explained in relative energy absorption at the knee (p=0.73). Once accounting for Tanner stage, strength, body composition, and cardiovascular fitness were not significant predictors of relative energy absorption at the knee (R^2 change = 6.0%, p=0.42). The final model can be found in Table 8.

Table 8. Regression summary results for each dependent variable prior to exercise.

Variable	R^2 (P value)	R^2 Change (P value)	Unstandardized Coefficients				
			Constant	Tanner stage	Broad Jump	VO^2	Body Fat
Initial Valgus	.039 (.17)		-2.32	.99			
	.068 (.52)	.030 (.70)	-1.64	1.22	.02	-.04	-.12
Peak Valgus	.003 (.70)		-7.56	-.47			
	.120 (.21)	.117 (.13)	-7.11	.13	.09	-.21	-.38
Valgus Excursion	.048 (.13)		-5.25	-1.46			
	.146 (.12)	.098 (.18)	-5.47	-1.09	.07	-.17	-.26
Initial Internal Rotation	.000 (.94)		-4.21	.12			
	.052 (.65)	.052 (.49)	-15.13	.25	.09	-.01	-.15
Peak Internal Rotation	.017 (.37)		3.94	1.27			
	.204 (.03)*	.187 (.02)*	-18.29	1.27	.18†	-.04	-.19
Internal Rotation Excursion	.032 (.21)		7.28	1.46			
	.103 (.29)	.071 (.33)	-5.61	1.27	.11	-.05	-.03
Relative EA	.003 (.73)		.371	-.009			
	.063 (.56)	.060 (.42)	.112	-.008	.002	.001	-.002

*Significant model (P<0.05)

† Significant regression coefficient (P<0.05)

Hypothesis 2: Contribution of Physical Measures to Landing Biomechanics after an Exercise Challenge

The following results describe the relationship between physical measures of strength, body composition, and fitness once accounting for Tanner stage on lower extremity biomechanics when landing biomechanics were measured following exercise.

Knee Valgus

When predicting knee valgus after exercise, Tanner stage accounted for 8.2% (p=0.04), 0.0% (p=0.99), and 4.1% (p=0.16) of variance explained in initial valgus, peak valgus, and valgus excursion, respectively. For every unit increase in Tanner stage, participants had 1.6 degrees less initial knee valgus when measured after exercise (as knee valgus is a negative value). Once accounting for Tanner stage, physical measures of strength, body composition, and cardiovascular fitness were not significant predictors of initial valgus (R^2 change=3.1%, p=0.67), peak valgus (R^2 change=10.0%, p=0.19), or valgus excursion (R^2 change=8.0%, p=0.27). The final models can be found in Table 9.

Tibial Rotation

When predicting internal tibial rotation after exercise, Tanner stage accounted for 0.5% (p=0.64), 0.8% (p=0.53), and 0.4% (p=0.65) of variance explained in initial internal tibial rotation, peak internal tibial rotation, and internal tibial rotation excursion, respectively. Once accounting for Tanner stage, physical measures of strength, body composition, and cardiovascular fitness did not significantly increase the predicted variance in initial tibial rotation (R^2 change = 10.2%, p=0.18), peak tibial rotation (R^2 change = 15.0%, p=.06) and rotation excursion (R^2 change = 1.8%, p=0.85). However, while the full model for peak internal rotation was not significant, the coefficient for

broad jump distance was significant (p=0.03), once controlling for all other variables (Table 9). Similar to before exercise, 1 cm greater broad jump distance predicted a 0.17° greater peak internal tibial rotation.

Relative Knee Energy Absorption

When predicting relative knee energy absorption after exercise, and once accounting for Tanner stage (R^2=1.8%, p=0.36), physical measures of strength, body composition, and cardiovascular fitness did not significantly increase the predicted variance (R^2 change = 0.3%, p=0.99). The final model can be found in Table 9.

Table 9. Regression summary results for each dependent variable following exercise.

Variable	R^2 (P value)	R^2 Change (P value)	Unstandardized Coefficients				
			Constant	Tanner stage	Broad Jump	VO^2	Body Fat
Initial Valgus	.082 (.04)		-3.50	1.40			
	.113 (.24)	.031 (.67)	-3.26	1.58†	.03	-.06	-.11
Peak Valgus	.000 (.99)		-9.12	-.01			
	.100 (.31)	.100 (.19)	-13.49	.54	.08	-.06	-.31
Valgus Excursion	.041 (.16)		-5.62	-1.41			
	.121 (.20)	.080 (.27)	-10.22	-1.04	.05	.01	-.20
Initial Internal Rotation	.005 (.64)		-7.67	.73			
	.107 (.27)	.102 (.18)	-18.26	1.11	.13	-.10	-.30
Peak Internal Rotation	.008 (.53)		5.18	.99			
	.159 (.09)	.150 (.06)	-7.64	1.33	.17†	-.18	-.34
Internal Rotation Excursion	.004 (.65)		11.97	.56			
	.022 (.91)	.018 (.85)	8.03	.43	.06	-.10	-.02
Relative EA	.018 (.36)		.242	.024			
	.020 (.92)	.003 (.92)	.272	.027	.000	.000	-.001

† Significant regression coefficient (P<0.05)

Hypothesis 3: Influence of Exercise on Contribution of Physical Measures to Landing Biomechanics

The full models for all dependent variables prior to and following the exercise challenge are in Table 10. When comparing the strength of the coefficients before and after exercise, significant coefficients were observed for standing broad jump with peak internal tibial rotation. When descriptively examining the pre- and post- exercise models for peak internal tibial rotation, the coefficient for standing broad jump was essentially unchanged (0.18 vs. 0.17). Specifically, once accounting for all other independent variables in the model, an increase of broad jump by 1 cm results in a .18° increase in peak internal tibial rotation prior to and exercise, and .17° increase following exercise. In more clinically relevant terms, a participant who jumped 165 cm would have 1.8° greater peak internal tibial rotation prior to exercise, and 1.7° greater peak internal tibial rotation following exercise than a participant that jump 155 cm (a 10 cm greater distance).

In one case, initial knee valgus, an independent variable that was not significant prior to exercise became significant following exercise. Specifically, Tanner stage was a significant predictor of initial knee valgus following exercise (coefficient 1.58, $p=0.04$) indicating that for every one unit greater Tanner stage, initial valgus angle was predicted to be 1.58° less.

Table 10. Comparison of Full regression models before and after exercise

Dependent Variable		R^2 value (P value)	Final Regression Equation
Initial Knee Valgus	before exercise	0.068 (0.52)	$Valgus_{IN}$ = -1.64 + 1.22(Tanner) + 0.02(BJ) - 0.04(VO_2) - 0.12(BF)
	after exercise	0.113 (0.24)	$Valgus_{IN}$ = -3.26 + 1.58(Tanner)† + 0.03(BJ) - 0.06(VO_2) - 0.11(BF)
Peak Knee Valgus	before exercise	0.120 (0.21)	$Valgus_{PK}$ = -7.11 – .13(Tanner) + 0.09(BJ) - 0.21(VO_2) – 0.38(BF)
	after exercise	0.100 (0.31)	$Valgus_{PK}$ = -13.49 + 0.54(Tanner) + 0.08(BJ) - 0.05(VO_2) – 0.31(BF)
Valgus Excursion	before exercise	0.146 (0.12)	$Valgus_{EX}$ = -5.47 – 1.09(Tanner) + 0.07(BJ) - 0.17(VO_2) – 0.26(BF)
	after exercise	0.121 (0.20)	$Valgus_{EX}$ = -10.22 – 1.04(Tanner) + 0.05(BJ) + 0.01(VO_2) – 0.20(BF)
Initial Internal Rotation	before exercise	0.052 (0.65)	$Rotation_{IN}$ = -15.13 + 0.25(Tanner) + 0.09(BJ) - 0.01(VO_2) - 0.15(BF)
	after exercise	0.107 (0.27)	$Rotation_{IN}$ = -18.26 + 1.11(Tanner) + 0.13(BJ) - 0.10(VO_2) – 0.30(BF)
Peak Internal Rotation	before exercise	0.204 (0.03)*	$Rotation_{PK}$ = -18.29 + 1.27(Tanner) + 0.18(BJ)† - 0.04(VO_2) - 0.19(BF)
	after exercise	0.159 (0.09)	$Rotation_{PK}$ = -7.64 + 1.33(Tanner) + 0.17(BJ)† - 0.18(VO_2) - 0.34(BF)
Internal Rotation Excursion	before exercise	0.103 (0.29)	$Rotation_{EX}$ = -5.61 + 1.26(Tanner) + 0.11(BJ) - 0.05(VO_2) - 0.03(BF)
	after exercise	0.022 (0.91)	$Rotation_{EX}$ = 8.03 + 0.43(Tanner) + 0.06(BJ) - 0.10(VO_2) - 0.02(BF)
Relative Knee Absorption	before exercise	0.063 (0.56)	RelKEA = 0.112 - 0.008(Tanner) + 0.002(BJ) + 0.001(VO_2) - 0.002(BF)
	after exercise	0.020 (0.92)	RelKEA = 0.272 + 0.027(Tanner) - 0.000(BJ) + 0.000(VO_2) - 0.001(BF)

†Significant regression coefficient ($P<0.05$)

Exploratory multi-level models were also run to examine the direct effects of exercise on the variables of interest, though underpowered to identify interaction effects. All models are shown in Table 11. None of the interaction terms in the multi-level models were significant for knee valgus or rotation, indicating that the relationship between physical characteristics and knee kinematics did not significantly change following exercise.

For relative knee energy absorption, there was a significant Tanner stage * Exercise interaction (P=0.02) as well as Broad Jump * Exercise interaction (P=0.02). The significant interactions indicate that the relationship between relative knee energy absorption and both Tanner stage and broad jump was different following exercise than it was prior to exercise. In the case of Tanner stage, the coefficient increased following exercise by 0.03, indicating that Tanner stage was a stronger predictor of relative energy absorption after exercise. Specifically, for a one stage difference in Tanner stage, relative energy absorption at the knee was 3% greater when measured after exercise. For broad jump, the opposite relationship was observed, with the coefficient (i.e. strength of the association) decreasing by 0.002 following exercise.

Table 11. Multi-Level Model of Effect of Exercise and Physical Characteristics on all dependent variables

	Unstandardized Coefficients (P value)									
Variable	**Intercept**	**Tanner stage**	**Body Fat**	**Broad Jump**	**VO^2**	**Exercise**	**Tanner * Exercise**	**Body Fat * Exercise**	**Broad Jump * Exercise**	**VO^2 * Exercise**
Initial Valgus	-1.64 (0.83)	1.22 (0.12)	-0.12 (0.33)	0.02 (0.61)	-0.04 (0.74)	-1.62 (0.72)	0.36 (0.44)	0.01 (0.89)	0.01 (0.74)	-0.02 (0.78)
Peak Valgus	-7.11 (0.58)	0.13 (0.92)	-0.38 (0.08)	0.09 (0.16)	-0.21 (0.34)	-6.37 (0.50)	0.42 (0.67)	0.06 (0.68)	-0.01 (0.80)	0.16 (0.33)
Valgus Excursion	-5.47 (0.59)	-1.09 (0.29)	-0.25 (0.13)	0.07 (0.16)	-0.17 (0.32)	-4.75 (0.54)	0.06 (0.95)	0.05 (0.68)	-0.02 (0.62)	0.18 (0.18)
Initial Internal Rotation	-15.13 (0.35)	0.25 (0.88)	-0.15 (0.58)	0.09 (0.23)	-0.01 (0.97)	-3.13 (0.65)	0.86 (0.65)	-0.16 (0.18)	0.03 (0.30)	-0.09 (0.46)
Peak Internal Rotation	-18.29 (0.22)	1.27 (0.40)	-0.19 (0.44)	0.18 (0.01)†	-0.04 (0.89)	10.65 (0.16)	0.06 (0.94)	-0.15 (0.23)	-0.01 (0.69)	-0.15 (0.24)
Internal Rotation Excursion	-5.61 (0.66)	1.26 (0.34)	-0.03 (0.88)	0.10 (0.09)	-0.05 (0.83)	13.61 (0.13)	-0.83 (0.36)	0.01 (0.94)	-0.05 (0.24)	-0.06 (0.71)
Relative EA	0.112 (0.68)	-0.008 (0.78)	-0.002 (0.70)	0.002 (0.22)	0.001 (0.76)	0.160 (0.27)	0.035 (0.02)†	0.003 (0.90)	-0.002 (0.02)†	-0.002 (0.47)

†Significant regression coefficient ($P<0.05$)

CHAPTER V

DISCUSSION

The primary purpose of this study was to examine the extent that physical characteristics of strength, body composition and fitness, as assessed via field based measures, were associated with knee joint biomechanics during landing in adolescent females, after accounting for their maturation stage. These physical characteristics were chosen for their potential to be modified through appropriate training. Further, field based tests to represent these physical characteristics were specifically chosen to represent tests that can easily be performed by clinicians screening for injury risk. The primary findings were that a greater Tanner stage of maturation was related to less predicted initial knee valgus angle following exercise, while greater functional lower extremity strength as ,measured by the standing broad jump was related to greater predicted peak internal tibial rotation. There were no associations between physical characteristics and relative energy absorption at the knee. Furthermore, exercise had little to no effect on these associations. The following sections will discuss the findings of this investigation, and the potential clinical implications.

Associations of Physical Characteristics and Maturation with Knee Biomechanics

Around age 12, when sex differences in ACL injury rates begin to emerge (Gianotti, et al., 2009; Le Gall, et al., 2006; Peterson, et al., 2000), knee biomechanics in females trend toward higher risk strategies, which include increased knee valgus (Ford, et

al., 2003; Hewett, et al., 2004) and increased reliance on the knee extensor muscles to dissipate landing forces (Sigward, et al., 2011). This propensity for developing higher risk knee joint biomechanics is thought to largely result from changes in physical characteristics that also occur during this time (Ahmad, et al., 2006; Barber-Westin, et al., 2005; Barber-Westin, et al., 2009; Hass, et al., 2003; Quatman, et al., 2006). However, these relationships have rarely been examined directly, which was the focus of this study. As such we examined the independent effects of strength, body composition, and cardiovascular fitness on landing biomechanics, after controlling for maturation.

Strength

For the current investigation, a standing broad jump was chosen as a field based measure of lower extremity strength and functional capabilities of each individual. During practice and competition when injury events occur, athletes perform propulsive and landing/stopping tasks that require each athlete to utilize the strength capabilities each individual possesses to decelerate their mass. The standing broad jump reflects both strength and deceleration, as participants were required to propel themselves forward as far as they could, and control their landings by landing on two feet without taking steps forward or backward. The distribution of jumping distances (Mean: 151.49 cm SD: ±17.38 Range: 114.67 - 193.33) observed in this study is consistent with previously published normative data in similar adolescent female populations (Castro-Pinero et al., 2009). The lack of a statistical difference in jump distance across Tanner stages ($p=0.65$) is also consistent with previous research that did not find an increase in lower extremity

strength (normalized to body mass) after age 11 (Barber-Westin, et al., 2006; Buchanan & Vardaxis, 2003, 2009; Myer, et al., 2009).

Decreased strength has been implicated as a potential factor in the increased incidence of ACL injury in female athletes. Specifically, it has been proposed that decreased strength results in decreased stability about the joint and an inability to adequately dampen joint loads during dynamic motions (Li et al., 1999; Myer, et al., 2009), thus resulting in inappropriate force attenuation strategies (Hewett, et al., 2004; Quatman, et al., 2006). As such, this study sought to test this theory by assessing whether lower extremity functional strength was associated with at-risk landing biomechanics in a population that has previously been found to be at an increased risk for ACL injury (Csintalan, et al., 2008; Gianotti, et al., 2009; Le Gall, et al., 2006; Peterson, et al., 2000; Shea, et al., 2004). While functional strength was not found to be associated with knee valgus or relative energy absorption at the knee, it was predictive of peak internal tibial rotation. Specifically, for every 10 cm further that a participant jumped, peak internal tibial rotation was predicted to increase by 1.7° before exercise and 1.8° following exercise. As such, the findings were contrary to our hypotheses as increased strength was predictive of *increased* peak internal tibial rotation.

As the direction of our results related to peak internal tibial rotation were not expected, we examined whether increased strength was related to a different overall landing strategy. Specifically, we examined whether there were relationships between strength and initial, peak, and excursion angles in the sagittal plane. These variables were explored as a longer broad jump is accomplished by maximizing mechanical

energy, of which increased range of motion in the sagittal plane is a mechanism (Nagano, Komura, & Fukashiro, 2007). Specifically, increased hip flexion has been found to increase horizontal jumping distance (Nagano, et al., 2007). No statistically significant relationship was found between broad jump distance and initial knee flexion (R^2=0.023 before exercise, R^2=0.007 after exercise), peak knee flexion (R^2=0.060 before exercise, R^2=0.035 after exercise), or knee flexion excursion (R^2=0.038 before exercise, R^2=0.027 after exercise). However, broad jump distance was a significant predictor of peak hip flexion both before (R^2=0.133) and after exercise (R^2=0.120), though no statistically significant relationship was found for initial hip flexion (R^2=0.053 before exercise, R^2=0.028 after exercise) or hip flexion excursion (R^2=0.032 before exercise, R^2=0.045 after exercise). This indicates that while participants that jumped further were not predicted to land in a more knee flexed position or go through more knee flexion during landing, they were predicted to experience greater peak hip flexion during landing. As these participants likely also utilized greater hip flexion during the broad jump to propel themselves further, it can be speculated that this jumping strategy translated to landing as well.

Additional exploratory analyses were performed to explore the relationship between frontal and sagittal plane motion. Given previous work that found subjects with less sagittal plane motion to have greater frontal plane motion (Pollard, et al., 2010), the relationship between peak knee valgus and peak hip and knee flexion was also examined. However, peak sagittal plane motion was not statistically related to peak frontal plane motion in our study either before (R^2=.113) or after (R^2=.049) exercise. Therefore, while

participants that were able to jump further appear to alter their landing pattern by landing with greater peak hip flexion, this strategy did not translate to greater knee flexion, nor decreased knee valgus. Further discussion of the relationships between strength and landing biomechanics follow.

Strength and Internal Tibial Rotation

Though functional strength has not been examined relative to transverse plane mechanics in prior work, other measures of lower extremity strength have been. In one study where rotational strength of the knee was related to shank rotation in adult females during a single leg landing (Kiriyama, Sato, & Takahira, 2009), a negative relationship between peak internal tibial rotation and external rotation strength of the shank was observed in females but not males. This suggests that lower strength of tibial external rotators (bicep femoris) may influence internal rotation of the tibia, perhaps as a result of not being able to produce a sufficient external rotation moment to counteract the internal moment associated with landing (Kiriyama, et al., 2009). In the current study, the standing broad jump is a functional strength assessment and thus contributions of individual muscles cannot be isolated. As such, it is difficult to compare our results to previous research.

The roles of the various thigh muscles in controlling dynamic tasks such as the broad jump will be discussed in more detail later, briefly however, hip extension occurs as a result of both gluteus maximus and three muscles of the hamstring muscle group (Drake, Vogl, Mitchell, & Gray, 2010). One component of the hamstring muscle group, the short head of the bicep femoris, is not involved with this muscle action (Drake, et al.,

2010). As such, it is possible that those who jump further have greater tibial internal rotation strength (semimembranosus and semitendinosus) than those who do not jump as far, since both internal tibial rotators are also hip extensors. As females were also found to have a significantly lower external rotation:internal rotation strength ratio than males (Kiriyama, et al., 2009), perhaps any increases in internal rotation strength further limit the ability of the external rotators to resist the internal rotation moment.

Strength and Knee Valgus

More work has been done examining the association between strength and frontal plane motion, with lower extremity strength previously related to knee valgus and medial displacement during landing (Barber-Westin, et al., 2006; R. J. Schmitz, et al., 2009; Wild, Steele, & Munro, 2013a). Of these studies, one used a functional strength assessment (R. J. Schmitz, et al., 2009) while the others used isokinetic dynamometers (Barber-Westin, et al., 2005; Barber-Westin, et al., 2006; Wild, et al., 2013a). While the statistical approaches, groupings, and strength assessments differed between the studies, the majority of the studies did not find a relationship between strength and frontal plane knee motion (Barber-Westin, et al., 2005; Barber-Westin, et al., 2006; R. J. Schmitz, et al., 2009). Specific to functional strength, triple hop distance was not able to predict dynamic valgus by maturation or sex (R^2 range: 0.005-0.084) (R. J. Schmitz, et al., 2009). The current study found a similar relationship between the standing broad jump and valgus excursion (comparable to the dynamic valgus measure used by Schmitz and colleagues) with an R^2 of 0.039 prior to exercise and 0.031 following exercise when

entered as the sole predictor. Together, this suggests that functional jumping tasks may not be related to knee valgus, either alone or once accounting for maturation stage.

However, a recent study did find that adolescent females with lower hamstring strength land in a greater knee valgus position. In this approach, groups were split based on high (33% that produced the greatest hamstring torque; n=11) and low (33% that produced the least hamstring torque; n=11) strength as measured via isokinetics. Females aged 10 -13 with lower hamstring strength landed with greater knee valgus at peak vertical and anterior/posterior GRF (Wild, et al., 2013a). It should be noted that in addition to the differing statistical approach, the landing task used to assess landing biomechanics in the previously mentioned study differed from that used in the current investigation, as well as the studies that did not find a relationship between functional strength and frontal plane motion. The task Wild and colleagues (Wild, et al., 2013a) used was a horizontal leap where subjects would take off from two feet, perform a maximum horizontal jump, and land on a single limb. Despite the group differences in hamstring torque production, there was no difference in the jump distance between the groups (Wild, et al., 2013a). The lack of difference in jump capabilities may indicate that the functional strength assessment of standing broad jump used in the current study may not be sensitive enough to distinguish between deficits in hamstring torque producing capability. This may be important as maturing females have been found to increase quadriceps strength while hamstring strength plateaus (Buchanan & Vardaxis, 2003), and this imbalance may not be picked up in a standing broad jump.

A key point to make regarding the relationship between broad jump distance and landing biomechanics is the strength of the relationships. Specifically, the relationship between broad jump distance and peak internal tibial rotation is statistically significant, with the model explaining an additional 18.7% of variance beyond Tanner stage prior to exercise and 15% following exercise. As such, the relationship is not strong, though may be described as moderate. For all other variables the relationship between broad jump distance and landing biomechanics is weak. Thus, while strength is moderately related to peak internal tibial rotation in a group of maturing females, other factors appear to contribute to landing biomechanics as well and should be considered for exploration in future studies. Other possible contributing factors will be discussed in later sections.

Muscle Contributions and Standing Broad Jump

Given the closed chained, functional nature of the standing broad jump, attention should be given to the multiple structures that are involved in successfully completing the task. Considering the contribution of muscle groups to the standing broad jump, the propulsion of the body is a result of ankle plantarflexion, knee extension, and hip extension. Of those components the hamstring muscle group would only assist with hip extension. Hip extension occurs as a result of gluteus maximus, semimembranosus, semitendinosus, and long head of the bicep femoris muscle contraction (Drake, et al., 2010), with different firing patterns and amplitude of muscle activity of these muscles based on hip abduction position (Kang, Jeon, Kwon, Cynn, & Choi, 2013). For example, as the angle of hip abduction increases, EMG amplitude of the gluteus maximus has also been shown to increase, while EMG amplitude of the hamstring muscle group decreases.

Likewise, relative onset of muscle activation has been shown to move toward the gluteus maximus as hip abduction angle increases (Kang, et al., 2013). As such, without knowing hip position during propulsion of the broad jump it is difficult to distinguish relative amount of hamstring or gluteus maximus influence on hip extension and therefore its contribution to the standing broad jump outcome.

Also worth considering is the contribution of the hamstring group as biarticular muscles during the jumping task. The relative contribution of the hamstrings to hip work during jumping and pushing off (as with a sprint start) has been reported to be between 7-11% (Jacobs, Bobbert, & van Ingen Schenau, 1996). When considered along with reports of the hamstrings to have a negative work output during horizontal jumping tasks (Nagano, et al., 2007), it appears that the role of the hamstrings in the broad jump may be an eccentric action in which the muscle stretches while exerting a force. While the hamstrings are believed to act eccentrically during a drop jump landing to assist with hip flexion and forward motion of the trunk (Devita & Skelly, 1992) as well as to assist with loading the hip for the subsequent vertical jump (Lees, Vanrenterghem, & De Clercq, 2004), the previously mentioned relative contribution of this muscle group to hip work is likely related to the lack of a statistical relationship between broad jump distance and energy absorption strategies. This further supports the position that while the broad jump is a valid measure of lower extremity functional strength (Castro-Pinero, Ortega, et al., 2010; Ortega, et al., 2008), it does not appear capable of differentiating potential muscle imbalances within or between the muscle groups of the lower extremity, and thereby why

we may not have seen clear relationships between strength and frontal plane motion as others have found (Wild, et al., 2013a).

Summary

The functional task of a standing broad jump was used as an assessment of each participant's ability to propel and control their body. Lower extremity strength was not found to be different in participants from different maturation stages, nor was it statistically related to frontal plane biomechanics at the knee or relative energy absorption at the knee. However, greater standing broad jump distance was related to greater peak internal tibial rotation both prior to and following exercise, though the strength of the relationship did not differ with exercise. As adolescent females with different hamstring torque production capability have not been found to have different functional strength capabilities (Wild, et al., 2013a), it is likely that this jumping task does not have the ability to differentiate between contribution of specific muscle groups.

Body Composition

Body composition may also play a role in divergent changes in landing biomechanics and injury risk during adolescence, in that females increase body fat during this time (Heyward & Wagner, 2004; Loomba-Albrecht & Styne, 2009), and an increased BMI has been linked to multiple injuries in adolescents (Bazelmans, et al., 2004; Doan, et al., 2010). Additionally, differences in body composition during maturation have been proposed as a factor in the emergence of more at-risk landing biomechanics (Ford, Shapiro, et al., 2010; Quatman, et al., 2006). Specific mechanisms that support this proposal is that the increased weight relative to lean body mass in those with greater

percent body fat increases the demand on the muscles of the lower extremity to control the additional weight during landing maneuvers (Montgomery, et al., 2012). However, in the current study we observed no direct relationship between landing biomechanics and percent body fat measured via skinfold thickness, despite a wide range in percent body fat (10.7% - 38.5%) (Table 5) and landing biomechanics (Table 6) across participants. This, in conjunction with the strength findings, suggests that females have sufficient muscle mass to safely decelerate their body during landing.

As such, the results of this study do not support the prevailing theory that females are at a biomechanical disadvantage in dissipating landing forces simply due to their body composition. However, it is acknowledged that this was a single sex study, thus the findings are based on normal body composition ranges in adolescent females, confirmed by comparison to normative data (Loomba-Albrecht & Styne, 2009). It should be noted however, that though the Slaughter equations have been commonly used in studies involving children and adolescents (Freedman, et al., 2007; Rodriguez, et al., 2005; Steinberger, et al., 2005), demonstrate relatively low error and bias (Rodriguez, et al., 2005), and have been recommended for use in clinical settings (Rodriguez, et al., 2005), there is inherent measurement error in skinfold thickness assessment. As fat distribution changes in adolescent females, specifically increasing in the waist and hips (gluteo-femoral region) (Loomba-Albrecht & Styne, 2009), the two upper extremity measurement sites may underestimate body fat particularly in the participants above Tanner stage 3.

While body composition changes throughout maturation (Heyward & Wagner, 2004; Loomba-Albrecht & Styne, 2009), as do landing biomechanics (Hewett, et al., 2004; R. J. Schmitz, et al., 2009) the lack of statistical relationship between the two in this study suggest that other factors influence biomechanics and should be considered in the future. Discussion of other potential factors will be discussed in following sections.

Cardiovascular Fitness

Cardiovascular fitness was included as a physical characteristic that may be related to landing biomechanics as it has been reported to decrease with maturation (Janz, et al., 2000; McMurray, et al., 2003), and it influences resistance to fatigue via ability to work at an increased submaximal VO_2 for longer periods of time (Hoogeveen, 2000). Resistance to fatigue is important as injury rates increase later in practice and competition in adolescents (Price, et al., 2004), and biomechanics have previously been shown to change following a fatiguing protocol in adults (Borotikar, et al., 2008; Sanna & O'Connor, 2008).

Contrary to the proposed hypotheses, estimated VO_{2max} was not associated with landing biomechanics. Additionally, in the current study estimated VO_{2max} did not appear to decrease in those in a higher stage of Tanner maturation (R^2=0.005, P=0.62), nor were there statistical differences in cardiovascular fitness between participants of different Tanner stage (P=0.41). When examining data regarding type and intensity of training, there was also no statistically significant relationship between activity level and maturation (P=0.72). This indicates that participants in this study were similarly active

regardless of maturation level, therefore it can be postulated that level of training equally affected participants.

As estimated VO_{2max} has previously been shown to decrease with maturation, the lack of decline at higher maturation levels was unexpected. The most plausible explanation for this finding is the similar activity level of all participants. As discussed previously, cardiovascular fitness is determined by both genetics (An, et al., 2000) and level of training (Hoogeveen, 2000). Thus the level of activity in the current study may be different from those investigated in previous works that have found a decrease in cardiovascular fitness throughout maturation. For example, in two studies that tracked both physical fitness and activity longitudinally for at least five years, a decline in peak VO_2 was accompanied by a decline in physical activity (McMurray, et al., 2003) though the other did not observe a decline in physical activity (Janz, et al., 2000). Inclusion of physical activity data in longitudinal studies that also tracks cardiovascular fitness should be considered in the future to determine the mechanism and extent of potential cardiovascular fitness decline in maturing athletes.

Maturation Level

The majority of the previous research done on adolescents that has grouped participants by maturation found a maturation effect for landing biomechanics, particularly knee valgus or medial displacement (Ford, et al., 2003; Hewett, et al., 2004; R. J. Schmitz, et al., 2009). As such, this investigation controlled for maturation level prior to examining the independent effects of the physical measures. When run in this fashion, the current investigation also revealed a statistically significant relationship for

knee biomechanics with level of maturation where Tanner stage was the lone predictor of initial knee valgus. However, greater Tanner stages were associated with less initial knee valgus. This relationship was opposite of what was expected as previous work has found knee valgus to increase with maturation. Moreover, no statistically significant relationship was found between Tanner stage and transverse plane motion.

The direction of the relationship between level of maturation and initial knee valgus was perhaps the most surprising result of this study. While the theory that greater maturation level would be related to greater knee valgus was based on previous work that found females to increase their valgus positioning during landing at higher maturation levels (Ford, et al., 2003; Ford, Shapiro, et al., 2010; Hewett, et al., 2004; R. J. Schmitz, et al., 2009), not all literature is in agreement with those findings. Specifically, no difference has been reported in knee separation distance when grouping participants by age (Barber-Westin, et al., 2006), and no difference in knee abduction ROM was observed between prepubescent and postpubescent participants (Hass, et al., 2005). Thus the relationship between frontal plane motion and age or maturation is mixed at best. Moreover, the findings of the current study are in closest agreement to a previous study where females have been found to land in a more varus position with age (though not examined by maturation) and go through more valgus motion during a stop jump task (Yu, et al., 2005). As such, it is worth acknowledging that perhaps the relationship between maturation and landing biomechanics is not as straightforward as previously proposed and other factors must be considered to fully understand this relationship.

Other factors that may influence these relationships or yield inconsistent findings are the range in age / maturation level studied, and the methods by which frontal plane knee motion was derived. Regarding the latter, most studies examining frontal plane motion across maturation groups used a 2D analysis. Two dimensional frontal plane measures previously used (R. J. Schmitz, et al., 2009) are not a measure of frontal plane arthrokinematic rotation, but rather lower extremity function in all three planes. Likewise, coronal displacement, even when measured from three dimensional motion capture data (Ford, et al., 2003; Hewett, et al., 2004), is not a true measure of frontal plane angle. Further complicating direct comparisons are differing methods of assessing maturation (Hewett, et al., 2004) along with varying participant age ranges. Participants as young as 8 (Hass, et al., 2005) and as old as 18 (R. J. Schmitz, et al., 2009) have previously been included, and likely incorporated a larger range of Tanner maturation levels than what was captured in the current study (ages 11-15, and limited to Tanner stages ranging from 1.5 – 4.5).

As such, worth considering as a factor that may have influenced the relationships may be the limited range in Tanner stages of maturation participants in this study, as pre pubertal (<1.5) and adult levels (> 4.5) of sexual maturation were not represented. Lacking participants in these extreme ranges may have influenced the results as peak height velocity (growth) occurs during Tanner stage 3, while linear growth begins during stage 2 and ends at stage 5 (Barnes, 1975; Tanner, 1962). Temporally following the linear growth spurt by 3 to 6 months is peak weight gain (Barnes, 1975). Not including participants in stages of stable growth and development may impact the ability to identify

clear linear relationships between variables, considering the variable and rapid changes occurring during mid-maturation stages.

Other factors that may contribute to inconsistent relationships are other variables not included in the analyses but that could combine with maturation to differentially influence landing biomechanics. One potential confounding factor is that participants' lower extremity anatomical alignment. Previous investigations have found that both Q angle and tibiofemoral angle are greater in women than men, and increase in women with maturation (Nguyen & Shultz, 2007; Shultz, et al., 2008). As both of these measures are in the frontal plane, these anatomical alignment variables would influence initial knee valgus positioning (particularly in 2D frontal plane measures), but may not be as influential in gross transverse knee rotation measures used in this study.

Another variable potentially influencing maturation effects on frontal plane motion is knee laxity. In addition to strength and body composition, research has also shown anterior knee laxity changes with maturation (Quatman, Ford, Myer, Paterno, & Hewett, 2008; Shultz, et al., 2008; Wild, Steele, & Munro, 2013b), and can influence knee biomechanics at landing (Shultz & Schmitz, 2009; Shultz et al., 2012; Shultz, Schmitz, Nguyen, & Levine, 2010). Specifically, females with above average frontal and transverse plane laxity have been found to be positioned with greater hip adduction and knee valgus early in landing (Shultz & Schmitz, 2009) and those with higher anterior knee laxity absorb more energy about the knee (Shultz, Schmitz, Nguyen, & Levine, 2010). As the current study did not control for laxity, it is unknown whether participants in this study had above or below average laxity profiles and therefore suppositions on the

influence of their laxity profiles on landing biomechanics cannot be made. However, if the participants in this study had low laxity, that may have influenced their frontal plane motion to the extent that may have further explained the positioning at landing being contrary to the proposed hypotheses.

Also changing with sexual maturation, and potentially influencing physical characteristics such as strength, laxity, and body composition are hormones. Hormonal fluctuations, particularly an increase in estrogen level, are known to occur in females throughout puberty. Estrogen levels rise continually from Tanner stages 2 – 5, and reduce during adulthood (Malina, Bouchard, & Bar-Or, 2004). While repetitive loading of the ACL, similar to what is experienced during daily activities, has been postulated as a mechanism of maintaining the integrity of the ACL (Toyoda, Matsumoto, Fujikawa, Saito, & Inoue, 1998), downregulation of collagen mRNA expression has been shown in porcine ACLs when subjected to cyclical loading while in an estrogen environment (Lee et al., 2004). This suggests that exposure to estrogen may decrease the strength of ACL, potentially negating the positive loading effects. Additionally, ACL of rats exposed to high estrogen environments (similar to rates experienced during puberty) absorbed less energy prior to failure, and showed lower deformation to failure (Woodhouse et al., 2007). Collectively, these results suggest that the hormonal changes experienced during maturation may affect the mechanical properties, and subsequently the laxity, of the ACL. However, as with laxity, the current study did not control for estrogen level or for the time in the cycle when testing was performed. Given the potential effects of estrogen on laxity, and laxity's effects on landing biomechanics, hormone levels may have also

further explained the unexpected landing biomechanics as they relate to maturation. This would be particularly true if participants in this study had hormone levels outside the normative values for females in the included age and maturation range.

Collectively, there are a number of factors that change with maturation, but that may not coincide directly with maturation as timing of changes vary from person to person, that may have impacted the relationship between maturation and landing biomechanics. As such, and when considering the fact that the physical characteristics measured in this study did not have strong relationships with landing biomechanics, researchers need to consider that injury risk and at-risk biomechanics may not be as straightforward as previously proposed. As ACL injury is a multi-faceted problem in a population known to have wide variability in intrinsic and extrinsic properties with maturation, perhaps researchers need to take a more inclusive approach to examining the contributions these factors may play in injury risk. Simply put, it may not be maturation and physical characteristics alone, but a collection of these factors along with variables such as alignment, laxity, hormones, and genetics. While much of the recent clinically driven research has not included these variables because they cannot be modified through training, understanding their respective roles in landing biomechanics is warranted. Moreover, the methods by which knee biomechanics are obtained many also be important in order to elucidate these relationships.

Summary

The current study related the physical characteristics of strength, body composition and fitness, assessed via field based measures, to knee joint biomechanics

during landing in adolescent females. These physical characteristics were chosen for their potential to be modified through appropriate training. The field based measures were chosen for their potential to be utilized in station based screening protocols, as well as to bring clinical meaning and ease of application to the measures. We found that a greater Tanner stage of maturation was related to less predicted initial knee valgus angle following exercise, and greater functional lower extremity strength was related to greater predicted peak internal tibial rotation angle. None of the included physical characteristics were statistically related to relative energy absorption at the knee. The lack of statistically significant relationships between the measures and landing biomechanics indicates that perhaps the relationship between these physical characteristics that change with maturation and landing biomechanics may not be as straightforward as previously proposed. As such, other factors must be considered to fully understand this relationship.

Effect of Exercise

Landing biomechanics were assessed both prior to and following exercise in this investigation in an attempt to extrapolate any potential negative effects of exercise, and by extension fatigue, on landing biomechanics. Effects of exercise on biomechanics were of interest for several reasons; first exercise has been found to negatively influence lower extremity biomechanics via increased knee valgus and rotation in adult female populations (Borotikar, et al., 2008; Sanna & O'Connor, 2008), however, to date this relationship has not been investigated in adolescents. Moreover, current injury screening investigations have either assessed landing mechanics prior to exercise or have not been clear when assessed (Goetschius et al., 2012; Myer, et al., 2012; Myer, et al., 2011;

Myer, et al., 2010; Smith, et al., 2012) even though it is well documented that injury rates are higher later in a practice or game (Price, et al., 2004). If exercise, and thus fatigue, effect landing biomechanics, this may have direct implications for the timing of the use of drop jumps as an injury screening mechanism.

A descriptive comparison of the models before and after the exercise challenge revealed statistical differences for a single outcome variable: initial valgus position. Prior to exercise, Tanner stage was not a significant predictor for initial knee valgus position, though following exercise it became significant. The coefficient for Tanner stage increased from 1.22 to 1.58 indicating that after exercise a participant in a higher Tanner stage was predicted to have 1.6° less knee valgus, while prior to exercise they were predicted to have 1.2° less knee valgus. But, while this relationship became somewhat stronger following exercise, the clinical implication of this increase appears to be negligible. Similarly, while broad jump was a significant predictor of peak internal tibial rotation both prior to and following exercise, the effect was similar pre- (β=0.18) and post- (β=0.17) exercise. These findings suggest that there is little benefit gained in screening athletes for these relationships after an exercise challenge. However, these findings are limited to a high intensity exercise challenge of short duration (12-15 minutes) that may not be representative of neuromuscular changes that occur during more prolonged intermittent exercise. Thus future studies should examine more sport specific exercise challenges, to see if these relationships hold for other forms of exhaustive activity.

To further explore the direct effect of the exercise challenge on knee biomechanics (regardless of different physical measures) a repeated measures ANOVA compared knee biomechanics prior to and following exercise. These results did not reveal any changes in knee valgus measures (P range: 0.53 - 0.97) that are commonly assessed during injury screening (Goetschius, et al., 2012; Myer, et al., 2012; Myer, et al., 2011; Myer, et al., 2010; Padua, et al., 2009; Smith, et al., 2012). This lack of change in knee valgus held true when participants were analyzed together, and when grouped by maturation. While the exercise protocol in the current study differed from previous studies, the lack of an exercise effect on knee valgus positioning following a fatigue protocol is not uncommon (Cortes, Greska, Kollock, Ambegaonkar, & Onate, 2013; Cortes, Quammen, Lucci, Greska, & Onate, 2012; Kernozek, Torry, & Iwasaki, 2008; Quammen et al., 2012; Thomas, McLean, & Palmieri-Smith, 2010).

The exercise challenge used in the current study has previously been used and validated as a maximal performance test for aerobic fitness (Leger, et al., 1988; Ruiz, et al., 2008; Ruiz, et al., 2009), and as such likely fatigued participants beyond what is typical in a normal practice or competition, though perhaps via a different mechanism than the intermittent nature of practice or competition. Likewise, the drop jump is commonly used in injury risk screening in the pre-season (Myer, et al., 2012; Myer, et al., 2011; Myer, et al., 2010) and as a part of a team meeting (Goetschius, et al., 2012; Smith, et al., 2012). The clinical implications of the negligible effects of this type of exercise on knee valgus in the current study is that screening may be able to be completed either prior to or following a practice with similar results regarding knee valgus. This could be

beneficial in high school or club sports settings when a large number of athletes need to be screened in a relatively short period of time. This may be particularly important when injury screening may be composed of multiple stations of moderate activity, meaning athletes may be able to start and end in at various stations with limited alterations in frontal plane landing biomechanics. However, given the type of exercise challenge used in the current study, extrapolation to other types of exercise should be implemented with caution.

Contrary to our findings with knee valgus, initial tibial rotation (P=0.014) and internal rotation excursion (P=0.019) did change after the exercise challenge, with participants landing in a more externally rotated position (-3.80° vs. -5.17°) and going through greater internal rotation excursion (12.06° vs. 13.74°) after exercise. These finding support previous research in adult females that similarly found exercise induced a more external rotated landing position along with increased internal rotation (Borotikar, et al., 2008; McLean et al., 2007; McLean & Samorezov, 2009). As tibial rotation has been shown to increase stress on the ACL (Gabriel, et al., 2004; Kanamori, et al., 2000; Oh, et al., 2012) the effect of exercise on transverse plane mechanics may be more detrimental than frontal plane mechanics with respect to injury potential. Another potential explanation for the increase in transverse plane motion could be the effect of exercise on knee laxity. Knee laxity in all three planes has been found to increase during exercise (Shultz, Schmitz, Cone, Copple, et al., 2013), and in females greater anterior-posterior knee laxty has been associated with greater knee internal rotation (Shultz, Schmitz, Cone, Henson, et al., 2013)(in review). Thus, these changes may simply

represent the normal viscoelastic change in soft tissue with exercise (Nawata et al., 1999; Shultz, Schmitz, Cone, Copple, et al., 2013). As no physical characteristics measured via field tests were statistically related to these changes, that would suggest this biomechanical change (laxity) may be more predictive following exercise. As such, future studies should consider incorportating laxity measures to further explore the effects of both exercise and maturation on landing biomechanics.

Multi-level models used to further examine whether the coefficients were statistically different prior to and following the exercise challenge. The model revealed interaction effects for both Tanner stage *Exercise and Broad Jump*Exercise in relation to relative energy absorption. The statistically significant interactions indicate that Tanner stage predicted 3% greater relative energy absorption, and broad jump predicted 2% less relative energy absorption following exercise, once accounting for all other predictors in the model.

While the relationships between Tanner and Broad Jump distance as they related to relative knee energy absorption, changed prior to and following exercise, these relationships were not statistically significant when run as multiple linear regressions with all predictors, nor when run in a stepwise fashion. This indicates that while the relationship between both Tanner stage and relative energy absorption at the knee, and broad jump and relative energy absorption at the knee changed following exercise, they did not explain a significant amount of variance at either timepoint. Additionally, post hoc repeated measures analysis did not reveal a Tanner*Exercise effect for relative energy absorption (P=0.06). The lack of statistical effect with a repeated measures model

indicates that there is no exercise effect for Tanner stage alone, but rather the effect is reliant on the other predictors in the multi-level model. Thus, while the effects were statistically significant, the clinical implications of the interactions are negligible, further supporting the view that a drop jump can be used for screening purposes either prior to or following exercise.

Limitations and Future Directions

While the current study aids in gleaning insight into the effect of modifiable physical characteristics that change throughout maturation on at-risk landing biomechanics, there are several limitations. While field based measures that are clinician friendly and can easily be replicated are important for transference of results from the laboratory into a clinical setting, it appears that some of the measures may have lacked the specificity (ability to rule out a variable or condition) and sensitivity (ability to rule in a variable or condition) that may be needed to related the physical measures examined in this study to landing biomechanics. For example, while the standing broad jump is a valid and reliable test of lower extremity strength (Castro-Pinero, Ortega, et al., 2010; Ortega, et al., 2008) the literature suggests (as previously described) that the test does not have the ability to distinguish between the knee extension strength of the quadriceps and the hip extension strength of the hamstrings, or their relative balance. Also, given the potential limitations of the 2 site Slaughter equation discussion previously, future studies that use skinfold thickness as a body composition estimate should consider using additional sites, including suprailiac and thigh, and using adult equations for older

participants. Alternately, future investigations may consider the use of tetra-polar bioelectrical impedance (Rodriguez, et al., 2005).

Additionally, this study is limited to the field-based assessment of the three physical characteristics of functional strength, body composition, and cardiovascular fitness, which for reasons previously stated, may not be fully representative of all aspects of strength, body composition and fitness in the maturing females. Additionally, genetics, hormones, knee laxity, and postural alignment were not examined and may confound the relationships between physical characteristics and landing biomechanics. But, while more comprehensive models are needed to understand why biomechanics change with maturation, and other variables of strength, body composition and fitness should be considered as the results suggest that gross modifiable physical characteristics alone are likely not responsible for these biomechanical changes. Therefore simply screening on these measures may not be sufficient to identify adolescents likely to demonstrate at-risk landing strategies. This has important implications for our current prevention efforts that largely focus on modifiable characteristics through focus on balance and jumping/landing techniques.

As this was a cross-sectional study, differences in training cannot be ruled out as a factor influencing landing patterns. Information from questionnaires and discussions with the participants indicated that 12 of the participants (24%) were currently involved in, or had previously undergone, formal jumping and landing training. For these 12, this training was either from a sports performance specialist, or a formal setting such as the Parisi Speed School. It is possible that this training may have influenced the landing

patterns in these participants. However, post hoc analysis comparing the initial valgus positioning before and after the exercise challenge did not reveal that these subjects landed differently (e.g. in a significantly more varus position (P=0.23)). Thus, this training did not appear to influence the relationships found with initial knee valgus angle, as these participants were distributed across Tanner stages.

The present study used a cross-sectional approach to examine the effects of physical measures on landing biomechanics in an adolescent female population. This design was chosen as a first step in understanding the relationships between physical characteristics that have previously been shown to change with maturation, and landing biomechanics in representative group of adolescent females before moving on to a more costly and time intensive longitudinal study. However, to further explore the effects of changing physical measures on landing strategies throughout maturation, a longitudinal research design is warranted. A longitudinal study would allow evaluation of the changes within a person over time, and perhaps be more informative regarding the concomitant physical and biomechanical changes. Additional measures of physical characteristics that are known to change throughout maturation, such as lower extremity alignment, should also be included as they may influence changing landing biomechanics. Also warranted may be the inclusion of additional lower extremity functional strength measures as a way of attempting to isolate muscle groups more effectively.

While the dynamic landing task used in this study is often utilized in both injury screening protocols and research examining lower extremity biomechanics during landing, a drop jump task by nature limits horizontal movement. However, we chose this

task for the current study to most closely relate to the field based landing assessments commonly used (Goetschius, et al., 2012; Myer, et al., 2012; Myer, et al., 2011; Myer, et al., 2010; Smith, et al., 2012), and to be able to compare with previous assessments of landing biomechanics in this population. As ACL injury has been shown to often occur during change of direction and stopping maneuvers associated with sport, a more activity specific task may be warranted for inclusion in future studies. Examples of currently investigated maneuvers include a horizontal leap (Wild, et al., 2013a), and sidestep cutting (Golden, Pavol, & Hoffman, 2009; Kipp, McLean, & Palmieri-Smith, 2011; McLean, et al., 2005; McLean, Lipfert, et al., 2004; McLean, et al., 1999; Pollard, Sigward, Ota, Langford, & Powers, 2006; Sigward, et al., 2008; Sigward & Powers, 2006). Both maneuvers, because of their sport specific nature, should be considered as additional landing/deceleration tasks in future studies.

It is also important to acknowledge is that the relationships examined in the current study are limited to biomechanical outcomes during a double leg drop landing, and may not be associated with positioning at the time of injury (Koga, et al., 2010; Krosshaug, Nakamae, et al., 2007; Olsen, et al., 2004). While it may be possible that these variables may be related to future injury, no conclusions can be drawn regarding injury outcomes, or biomechanical outcomes associated with other functional tasks.

Finally, though adequately powered based on anticipated effect sizes, we did not capture participants representing all Tanner stages. Considering this limitation, coupled with the inherent measurement error of the field based measures previously addressed, a greater sample size may have yielded additional significant relationships. This

assumption is based on some models yielding R^2 values between 0.10 and 0.15 with p values less than 0.25 (initial valgus angle, peak valgus angle, valgus excursion, and initial internal tibial rotation angle). Future studies should consider a larger sample size that includes equal representation of participants in Tanner stages 1 and 5.

Summary and Conclusions

This investigation utilized field based assessments of physical measures of strength, body composition, and cardiovascular fitness as well as a Tanner self-assessment to investigate the relationships of these physical measures with high risk landing knee biomechanics associated with ACL injury. The field based assessments were chosen as a way of linking the laboratory measures to clinician friendly measures that have the potential to be used in team or school injury screening situations, and thus have potential for a greater potential impact on injury risk identification than laboratory measures that can be both more time demanding and expensive. An additional clinical benefit was the direct comparison of pre and post exercise landing biomechanics, again potentially influencing clinical practice by identifying the most appropriate state of rest or fatigue to perform drop jump. The findings were that Tanner stage of development alone accounted for 3.9% of variance explained in intial knee valgus position prior to exercise and 8.2% following exercise at which time it was a significant predictor. In both instances, greater knee *varus* positioning was related to greater Tanner stage of development. Additionally, standing broad jump distance was a statistically significant predictor of peak internal tibial rotation, both prior to and following exercise with greater broad jump distances being related to increased peak internal rotation of the tibia.

These relationships indicate that the field based measures of maturation and strength are related to landing biomechanics both before and after exercise. This relationship was further strengthened as interaction effects were found for both Tanner * Exercise and Broad Jump* Exercise in relation to relative energy absorption at the knee. Thus, it appears that landing mechanics that have been shown to change during adolescence cannot solely be attributed to potential changes in strength, body composition and fitness (as these variables never accounted for greater than 19% variance explained in any of the dependent variables) and additional factors should be included in future studies. However, each of these physical characteristic warrants further inclusion in future studies investigating changing landing biomechanics in populations of adolescent females that participate in athletics. Though only functional strength was statistically related to at-risk landing biomechanics in this representative population of adolescent females, the strength of the relationships with other variables suggests that with more subjects, particularly in Tanner stages 1 and 5, additional relationships may emerge.

REFERENCES

Addo, O. Y., & Himes, J. H. (2010). Reference curves for triceps and subscapular skinfold thicknesses in US children and adolescents. *Am J Clin Nutr, 91*(3), 635-642.

Ahmad, C. S., Clark, A. M., Heilmann, N., Schoeb, J. S., Gardner, T. R., & Levine, W. N. (2006). Effect of gender and maturity on quadriceps-to-hamstring strength ratio and anterior cruciate ligament laxity. *Am J Sports Med, 34*(3), 370-374.

An, P., Rice, T., Gagnon, J., Leon, A. S., Skinner, J. S., Bouchard, C., et al. (2000). Familial aggregation of stroke volume and cardiac output during submaximal exercise: the HERITAGE Family Study. *Int J Sports Med, 21*(8), 566-572.

Anderson, A. F., Dome, D. C., Gautam, S., Awh, M. H., & Rennirt, G. W. (2001). Correlation of anthropometric measurements, strength, anterior cruciate ligament size, and intercondylar notch characteristics to sex differences in anterior cruciate ligament tear rates. *Am J Sports Med, 29*(1), 58-66.

Arendt, E., & Dick, R. (1995). Knee Injury Patterns Among Men and Women in Collegiate Basketball and Soccer. *The American Journal of Sports Medicine, 23*(6), 694-701.

Arendt, E. A., Agel, J., & Dick, R. (1999). Anterior cruciate ligament injury patterns among collegiate men and women. *J Athl Train, 34*(2), 86-92.

Artero, E. G., Espana-Romero, V., Castro-Pinero, J., Ortega, F. B., Suni, J., Castillo-Garzon, M. J., et al. (2011). Reliability of field-based fitness tests in youth. *Int J Sports Med, 32*(3), 159-169.

Barber-Westin, S. D., Galloway, M., Noyes, F. R., Corbett, G., & Walsh, C. (2005). Assessment of lower limb neuromuscular control in prepubescent athletes. *Am J Sports Med, 33*(12), 1853-1860.

Barber-Westin, S. D., Noyes, F. R., & Galloway, M. (2006). Jump-land characteristics and muscle strength development in young athletes: a gender comparison of 1140 athletes 9 to 17 years of age. *Am J Sports Med, 34*(3), 375-384.

Barber-Westin, S. D., Noyes, F. R., Smith, S. T., & Campbell, T. M. (2009). Reducing the risk of noncontact anterior cruciate ligament injuries in the female athlete. *Phys Sportsmed, 37*(3), 49-61Barnes, H. V. (1975). Physical growth and development during puberty. *Med Clin North Am, 59*(6), 1305-1317.

Bazelmans, C., Coppieters, Y., Godin, I., Parent, F., Berghmans, L., Dramaix, M., et al. (2004). Is obesity associated with injuries among young people? *Eur J Epidemiol, 19*(11), 1037-1042.

Bell, A. L., Brand, R. A., & Pedersen, D. R. (1989). Prediction of Hip-Joint Center Location from External Landmarks. *Human Movement Science, 8*(1), 3-16.

Boden, B. P., Dean, G. S., Feagin, J. A., Jr., & Garrett, W. E., Jr. (2000). Mechanisms of anterior cruciate ligament injury. *Orthopedics, 23*(6), 573-578.

Boden, B. P., Torg, J. S., Knowles, S. B., & Hewett, T. E. (2009). Video analysis of anterior cruciate ligament injury: abnormalities in hip and ankle kinematics. *Am J Sports Med, 37*(2), 252-259.

Borotikar, B. S., Newcomer, R., Koppes, R., & McLean, S. G. (2008). Combined effects of fatigue and decision making on female lower limb landing postures: central and peripheral contributions to ACL injury risk. *Clin Biomech (Bristol, Avon), 23*(1), 81-92.

Bridge, P., Pocock, N. A., Nguyen, T., Munns, C., Cowell, C. T., Forwood, N., et al. (2011). Validation of longitudinal DXA changes in body composition from pre- to mid-adolescence using MRI as reference. *J Clin Densitom, 14*(3), 340-347.

Brophy, R. H., Wright, R. W., & Matava, M. J. (2009). Cost analysis of converting from single-bundle to double-bundle anterior cruciate ligament reconstruction. *Am J Sports Med, 37*(4), 683-687.

Buchanan, P. A., & Vardaxis, V. G. (2003). Sex-Related and Age-Related Differences in Knee Strength of Basketball Players Ages 11-17 Years. *J Athl Train, 38*(3), 231-237.

Buchanan, P. A., & Vardaxis, V. G. (2009). Lower-extremity strength profiles and gender-based classification of basketball players ages 9-22 years. *J Strength Cond Res, 23*(2), 406-419.

Butler, D. L., Grood, E. S., Noyes, F. R., & Zernicke, R. F. (1978). Biomechanics of ligaments and tendons. *Exerc Sport Sci Rev, 6*, 125-181.

Cahuzac, J. P., Vardon, D., & Sales de Gauzy, J. (1995). Development of the clinical tibiofemoral angle in normal adolescents. A study of 427 normal subjects from 10 to 16 years of age. *J Bone Joint Surg Br, 77*(5), 729-732.

Castro-Pinero, J., Artero, E. G., Espana-Romero, V., Ortega, F. B., Sjostrom, M., Suni, J., et al. (2010). Criterion-related validity of field-based fitness tests in youth: a systematic review. *Br J Sports Med, 44*(13), 934-943.

Castro-Pinero, J., Gonzalez-Montesinos, J. L., Mora, J., Keating, X. D., Girela-Rejon, M. J., Sjostrom, M., et al. (2009). Percentile values for muscular strength field tests in children aged 6 to 17 years: influence of weight status. *J Strength Cond Res, 23*(8), 2295-2310.

Castro-Pinero, J., Mora, J., Gonzalez-Montesinos, J. L., Sjostrom, M., & Ruiz, J. R. (2009). Criterion-related validity of the one-mile run/walk test in children aged 8-17 years. *J Sports Sci, 27*(4), 405-413.

Castro-Pinero, J., Ortega, F. B., Artero, E. G., Girela-Rejon, M. J., Mora, J., Sjostrom, M., et al. (2010). Assessing muscular strength in youth: usefulness of standing long jump as a general index of muscular fitness. *J Strength Cond Res, 24*(7), 1810-1817.

Castro-Pinero, J., Ortega, F. B., Mora, J., Sjostrom, M., & Ruiz, J. R. (2009). Criterion related validity of 1/2 mile run-walk test for estimating VO2peak in children aged 6-17 years. *Int J Sports Med, 30*(5), 366-371.

Chappell, J. D., Herman, D. C., Knight, B. S., Kirkendall, D. T., Garrett, W. E., & Yu, B. (2005). Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. *American Journal of Sports Medicine, 33*(7), 1022-1029.

Cohen, J. (1988). *Statistical power analysis for the behavioral sciences* (2nd ed.). Hillsdale, N.J.: L. Erlbaum Associates.

Cone, J. R., Berry, N. T., Goldfarb, A., Henson, R., Schmitz, R., Wideman, L., et al. (2011). Effects of an Individualized Soccer Match Simulation on Vertical Stiffness and Impedance. *J Strength Cond Res*.

Cortes, N., Greska, E., Kollock, R., Ambegaonkar, J., & Onate, J. A. (2013). Changes in lower extremity biomechanics due to a short-term fatigue protocol. *J Athl Train, 48*(3), 306-313.

Cortes, N., Quammen, D., Lucci, S., Greska, E., & Onate, J. (2012). A functional agility short-term fatigue protocol changes lower extremity mechanics. *J Sports Sci, 30*(8), 797-805.

Cowling, E. J., & Steele, J. R. (2001). Is lower limb muscle synchrony during landing affected by gender? Implications for variations in ACL injury rates. *Journal of Electromyography and Kinesiology, 11*, 263-268.

Csintalan, R. P., Inacio, M. C., & Funahashi, T. T. (2008). Incidence rate of anterior cruciate ligament reconstructions. *Perm J, 12*(3), 17-21.

Decker, M. J., Torry, M. R., Wyland, D. J., Sterett, W. I., & Steadman, J. R. (2003). Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clinical Biomechanics, 18*, 662-669.

Deitch, J. R., Starkey, C., Walters, S. L., & Moseley, J. B. (2006). Injury risk in professional basketball players: a comparison of Women's National Basketball Association and National Basketball Association athletes. *Am J Sports Med, 34*(7), 1077-1083.

Devita, P., & Skelly, W. A. (1992). Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Med Sci Sports Exerc, 24*(1), 108-115.

DiStefano, L. J., Blackburn, J. T., Marshall, S. W., Guskiewicz, K. M., Garrett, W. E., & Padua, D. A. (2011). Effects of an age-specific anterior cruciate ligament injury prevention program on lower extremity biomechanics in children. *Am J Sports Med, 39*(5), 949-957.

DiStefano, L. J., Padua, D. A., Blackburn, J. T., Garrett, W. E., Guskiewicz, K. M., & Marshall, S. W. (2010). Integrated injury prevention program improves balance and vertical jump height in children. *J Strength Cond Res, 24*(2), 332-342.

DiStefano, L. J., Padua, D. A., DiStefano, M. J., & Marshall, S. W. (2009). Influence of age, sex, technique, and exercise program on movement patterns after an anterior cruciate ligament injury prevention program in youth soccer players. *Am J Sports Med, 37*(3), 495-505.

Doan, Q., Koehoorn, M., & Kissoon, N. (2010). Body mass index and the risk of acute injury in adolescents. *Paediatr Child Health, 15*(6), 351-356.

Drake, R. L., Vogl, W., Mitchell, A. W. M., & Gray, H. (2010). *Gray's anatomy for students* (2nd ed.). Philadelphia, PA: Churchill Livingstone/Elsevier.

Dufek, J. S., & Bates, B. T. (1990). The evaluation and prediction of impact forces during landings. *Med Sci Sports Exerc, 22*(3), 370-377.

Evans, K. N., Kilcoyne, K. G., Dickens, J. F., Rue, J. P., Giuliani, J., Gwinn, D., et al. (2012). Predisposing risk factors for non-contact ACL injuries in military subjects. *Knee Surg Sports Traumatol Arthrosc, 20*(8), 1554-1559.

Fleming, B. C., Ohlen, G., Renstrom, P. A., Peura, G. D., Beynnon, B. D., & Badger, G. J. (2003). The effects of compressive load and knee joint torque on peak anterior cruciate ligament strains. *Am J Sports Med, 31*(5), 701-707.

Fleming, B. C., Renstrom, P. A., Beynnon, B. D., Engstrom, B., Peura, G. D., Badger, G. J., et al. (2001). The effect of weightbearing and external loading on anterior cruciate ligament strain. *J Biomech, 34*(2), 163-170.

Ford, K. R., Myer, G. D., Brent, J. L., & Hewett, T. E. (2009). Hip and knee extensor moments predict vertical jump height in adolescent girls. *J Strength Cond Res, 23*(4), 1327-1331.

Ford, K. R., Myer, G. D., & Hewett, T. E. (2003). Valgus knee motion during landing in high school female and male basketball players. *Med Sci Sports Exerc, 35*(10), 1745-1750.

Ford, K. R., Myer, G. D., & Hewett, T. E. (2010). Longitudinal effects of maturation on lower extremity joint stiffness in adolescent athletes. *Am J Sports Med, 38*(9), 1829-1837.

Ford, K. R., Myer, G. D., Toms, H. E., & Hewett, T. E. (2005). Gender differences in the kinematics of unanticipated cutting in young athletes. *Med Sci Sports Exerc, 37*(1), 124-129.

Ford, K. R., Shapiro, R., Myer, G. D., Van Den Bogert, A. J., & Hewett, T. E. (2010). Longitudinal sex differences during landing in knee abduction in young athletes. *Med Sci Sports Exerc, 42*(10), 1923-1931.

Freedman, D. S., Wang, J., Ogden, C. L., Thornton, J. C., Mei, Z., Pierson, R. N., et al. (2007). The prediction of body fatness by BMI and skinfold thicknesses among children and adolescents. *Ann Hum Biol, 34*(2), 183-194.

Gabriel, M. T., Wong, E. K., Woo, S. L., Yagi, M., & Debski, R. E. (2004). Distribution of in situ forces in the anterior cruciate ligament in response to rotatory loads. *J Orthop Res, 22*(1), 85-89.

Gehring, D., Melnyk, M., & Gollhofer, A. (2009). Gender and fatigue have influence on knee joint control strategies during landing. *Clin Biomech (Bristol, Avon), 24*(1), 82-87.

Gelber, A. C., Hochberg, M. C., Mead, L. A., Wang, N. Y., Wigley, F. M., & Klag, M. J. (2000). Joint injury in young adults and risk for subsequent knee and hip osteoarthritis. *Ann Intern Med, 133*(5), 321-328.

Gianotti, S. M., Marshall, S. W., Hume, P. A., & Bunt, L. (2009). Incidence of anterior cruciate ligament injury and other knee ligament injuries: a national population-based study. *J Sci Med Sport, 12*(6), 622-627.

Goetschius, J., Smith, H. C., Vacek, P. M., Holterman, L. A., Shultz, S. J., Tourville, T. W., et al. (2012). Application of a clinic-based algorithm as a tool to identify female athletes at risk for anterior cruciate ligament injury: a prospective cohort study with a nested, matched case-control analysis. *Am J Sports Med, 40*(9), 1978-1984.

Golden, G. M., Pavol, M. J., & Hoffman, M. A. (2009). Knee joint kinematics and kinetics during a lateral false-step maneuver. *J Athl Train, 44*(5), 503-510.

Gomez, E., DeLee, J. C., & Farney, W. C. (1996). Incidence of injury in Texas girls' high school basketball. *Am J Sports Med, 24*(5), 684-687.

Griffin, L. Y., Albohm, M. J., Arendt, E. A., Bahr, R., Beynnon, B. D., Demaio, M., et al. (2006). Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005. *Am J Sports Med, 34*(9), 1512-1532.

Grood, E. S., Suntay, W. J., Noyes, F. R., & Butler, D. L. (1984). Biomechanics of the knee-extension exercise. Effect of cutting the anterior cruciate ligament. *J Bone Joint Surg Am, 66*(5), 725-734.

Hamilton, R. T., Shultz, S. J., Schmitz, R. J., & Perrin, D. H. (2008). Triple-hop distance as a valid predictor of lower limb strength and power. *J Athl Train, 43*(2), 144-151.

Hass, C. J., Schick, E. A., Chow, J. W., Tillman, M. D., Brunt, D., & Cauraugh, J. H. (2003). Lower extremity biomechanics differ in prepubescent and postpubescent female athlets during stride jump landings. *J Appl Biomech, 19*, 139-152.

Hass, C. J., Schick, E. A., Tillman, M. D., Chow, J. W., Brunt, D., & Cauraugh, J. H. (2005). Knee biomechanics during landings: comparison of pre- and postpubescent females. *Med Sci Sports Exerc, 37*(1), 100-107.

Henry, J. C., & Kaeding, C. (2001). Neuromuscular differences between male and female athletes. *Curr Womens Health Rep, 1*(3), 241-244.

Hewett, T. E., Myer, G. D., & Ford, K. R. (2004). Decrease in neuromuscular control about the knee with maturation in female athletes. *J Bone Joint Surg Am, 86-A*(8), 1601-1608.

Hewett, T. E., Myer, G. D., & Ford, K. R. (2006). Anterior cruciate ligament injuries in female athletes: Part 1, mechanisms and risk factors. *Am J Sports Med, 34*(2), 299-311.

Hewett, T. E., Myer, G. D., Ford, K. R., & Slauterbeck, J. R. (2006). Preparticipation physical examination using a box drop vertical jump test in young athletes: the effects of puberty and sex. *Clin J Sport Med, 16*(4), 298-304.

Heyward, V. H., & Wagner, D. R. (2004). *Applied body composition assessment* (2nd ed.). Champaign, IL: Human Kinetics.

Hoogeveen, A. R. (2000). The effect of endurance training on the ventilatory response to exercise in elite cyclists. *Eur J Appl Physiol, 82*(1-2), 45-51.

Horita, T., Komi, P. V., Nicol, C., & Kyrolainen, H. (1996). Stretch shortening cycle fatigue: interactions among joint stiffness, reflex, and muscle mechanical performance in the drop jump [corrected]. *Eur J Appl Physiol Occup Physiol, 73*(5), 393-403.

Hughes, G., Watkins, J., & Owen, N. (2008). Gender differences in lower limb frontal plane kinematics during landing. *Sports Biomech, 7*(3), 333-341.

Hughes, G., Watkins, J., & Owen, N. (2010). The effects of opposition and gender on knee kinematics and ground reaction force during landing from volleyball block jumps. *Res Q Exerc Sport, 81*(4), 384-391.

Huston, L. J., Vibert, B., Ashton-Miller, J. A., & Wojtys, E. M. (2001). Gender differences in knee angle when landing from a drop-jump. *The Journal of Knee Surgery, 14*(4), 215-220.

Ireland, M. L. (1999). Anterior cruciate ligament injury in female athletes: Epidemiology. *Journal of Athletic Training, 34*(2), 150-154.

Jacobs, R., Bobbert, M. F., & van Ingen Schenau, G. J. (1996). Mechanical output from individual muscles during explosive leg extensions: the role of biarticular muscles. *J Biomech, 29*(4), 513-523.

James, C. R., Dufek, J. S., & Bates, B. T. (2006). Effects of stretch shortening cycle exercise fatigue on stress fracture injury risk during landing. *Res Q Exerc Sport, 77*(1), 1-13.

James, C. R., Sizer, P. S., Starch, D. W., Lockhart, T. E., & Slauterbeck, J. (2004). Gender differences among sagittal plane knee kinematic and ground reaction force characteristics during a rapid spring and cut maneuver. *Research Quarterlly for Exercise and Sport, 75*(1), 31-38.

Janz, K. F., Dawson, J. D., & Mahoney, L. T. (2000). Tracking physical fitness and physical activity from childhood to adolescence: the muscatine study. *Med Sci Sports Exerc, 32*(7), 1250-1257.

Kanamori, A., Woo, S. L., Ma, C. B., Zeminski, J., Rudy, T. W., Li, G., et al. (2000). The forces in the anterior cruciate ligament and knee kinematics during a simulated pivot shift test: A human cadaveric study using robotic technology. *Arthroscopy, 16*(6), 633-639.

Kang, S. Y., Jeon, H. S., Kwon, O., Cynn, H. S., & Choi, B. (2013). Activation of the gluteus maximus and hamstring muscles during prone hip extension with knee flexion in three hip abduction positions. *Man Ther*.

Kaul, S., Rothney, M. P., Peters, D. M., Wacker, W. K., Davis, C. E., Shapiro, M. D., et al. (2012). Dual-Energy X-Ray Absorptiometry for Quantification of Visceral Fat. *Obesity (Silver Spring)*.

Kemper, H. C., & Verschuur, R. (1987). Longitudinal study of maximal aerobic power in teenagers. *Ann Hum Biol, 14*(5), 435-444.

Kernozek, T. W., Torry, M. R., H, V. A. N. H., Cowley, H., & Tanner, S. (2005). Gender differences in frontal and sagittal plane biomechanics during drop landings. *Med Sci Sports Exerc, 37*(6), 1003-1012; discussion 1013.

Kernozek, T. W., Torry, M. R., & Iwasaki, M. (2008). Gender differences in lower extremity landing mechanics caused by neuromuscular fatigue. *Am J Sports Med, 36*(3), 554-565.

Kipp, K., McLean, S. G., & Palmieri-Smith, R. M. (2011). Patterns of hip flexion motion predict frontal and transverse plane knee torques during a single-leg land-and-cut maneuver. *Clin Biomech (Bristol, Avon), 26*(5), 504-508.

Kiriyama, S., Sato, H., & Takahira, N. (2009). Gender differences in rotation of the shank during single-legged drop landing and its relation to rotational muscle strength of the knee. *Am J Sports Med, 37*(1), 168-174.

Koga, H., Nakamae, A., Shima, Y., Iwasa, J., Myklebust, G., Engebretsen, L., et al. (2010). Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med, 38*(11), 2218-2225.

Krosshaug, T., & Bahr, R. (2005). A model-based image-matching technique for three-dimensional reconstruction of human motion from uncalibrated video sequences. *J Biomech, 38*(4), 919-929.

Krosshaug, T., Nakamae, A., Boden, B. P., Engebretsen, L., Smith, G., Slauterbeck, J. R., et al. (2007). Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. *Am J Sports Med, 35*(3), 359-367.

Krosshaug, T., Slauterbeck, J. R., Engebretsen, L., & Bahr, R. (2007). Biomechanical analysis of anterior cruciate ligament injury mechanisms: three-dimensional motion reconstruction from video sequences. *Scand J Med Sci Sports, 17*(5), 508-519.

Kuczmarski, R. J., Ogden, C. L., Guo, S. S., Grummer-Strawn, L. M., Flegal, K. M., Mei, Z., et al. (2002). 2000 CDC Growth Charts for the United States: methods and development. *Vital Health Stat 11*(246), 1-190.

Landry, S. C., McKean, K. A., Hubley-Kozey, C. L., Stanish, W. D., & Deluzio, K. J. (2007). Neuromuscular and lower limb biomechanical differences exist between male and female elite adolescent soccer players during an unanticipated run and crosscut maneuver. *Am J Sports Med, 35*(11), 1901-1911.

Le Gall, F., Carling, C., Reilly, T., Vandewalle, H., Church, J., & Rochcongar, P. (2006). Incidence of injuries in elite French youth soccer players: a 10-season study. *Am J Sports Med, 34*(6), 928-938.

Lee, C. Y., Liu, X., Smith, C. L., Zhang, X., Hsu, H. C., Wang, D. Y., et al. (2004). The combined regulation of estrogen and cyclic tension on fibroblast biosynthesis derived from anterior cruciate ligament. *Matrix Biol, 23*(5), 323-329.

Lees, A., Vanrenterghem, J., & De Clercq, D. (2004). The maximal and submaximal vertical jump: implications for strength and conditioning. *J Strength Cond Res, 18*(4), 787-791.

Leger, L. A., Mercier, D., Gadoury, C., & Lambert, J. (1988). The multistage 20 metre shuttle run test for aerobic fitness. *J Sports Sci, 6*(2), 93-101.

Leone, M., & Comtois, A. S. (2007). Validity and reliability of self-assessment of sexual maturity in elite adolescent athletes. *Journal of Sports Medicine and Physical Fitness, 47*(3), 361-365.

Li, G., Rudy, T. W., Allen, C., Sakane, M., & Woo, S. L. (1998). Effect of combined axial compressive and anterior tibial loads on in situ forces in the anterior cruciate ligament: a porcine study. *J Orthop Res, 16*(1), 122-127.

Li, G., Rudy, T. W., Sakane, M., Kanamori, A., Ma, C. B., & Woo, S. L. (1999). The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. *J Biomech, 32*(4), 395-400.

Liu, N. Y., Plowman, S. A., & Looney, M. A. (1992). The reliability and validity of the 20-meter shuttle test in American students 12 to 15 years old. *Res Q Exerc Sport, 63*(4), 360-365.

Lohman, T. G., Pollock, M. L., Slaughter, M. H., Brandon, L. J., & Boileau, R. A. (1984). Methodological factors and the prediction of body fat in female athletes. *Med Sci Sports Exerc, 16*(1), 92-96.

Lohmander, L. S., Ostenberg, A., Englund, M., & Roos, H. (2004). High prevalence of knee osteoarthritis, pain, and functional limitations in female soccer players twelve years after anterior cruciate ligament injury. *Arthritis Rheum, 50*(10), 3145-3152.

Loomba-Albrecht, L. A., & Styne, D. M. (2009). Effect of puberty on body composition. *Curr Opin Endocrinol Diabetes Obes, 16*(1), 10-15.

Malina, R. M., Bouchard, C., & Bar-Or, O. (2004). *Growth, maturation, and physical activity* (2nd ed.). Champaign, Ill: Human Kinetics.

Malinzak, R. A., Colby, S., Kirkendall, D., Yu, B., & Garrett, W. E. (2001). A comparison of knee joint motion patterns between men and women in selected athletic tasks. *Clinical Biomechanics, 16*, 438-445.

Markolf, K. L., Burchfield, D. M., Shapiro, M. M., Shepard, M. F., Finerman, G. A. M., & Slauterbeck, J. (1995). Combined knee loading states that generate high anterior cruciate ligament forces. *Journal of Orthopaedic Research, 13*, 930-935.

McLean, S. G., Fellin, R. E., Suedekum, N., Calabrese, G., Passerallo, A., & Joy, S. (2007). Impact of fatigue on gender-based high-risk landing strategies. *Med Sci Sports Exerc, 39*(3), 502-514.

McLean, S. G., Huang, X., Su, A., & Van Den Bogert, A. J. (2004). Sagittal plane biomechanics cannot injury the ACL during sidestep cutting. *Clinical Biomechanics, 19*, 828-838.

McLean, S. G., Huang, X., & van den Bogert, A. J. (2005). Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury. *Clin Biomech (Bristol, Avon), 20*(8), 863-870.

McLean, S. G., Lipfert, S. W., & Van Den Bogert, A. J. (2004). Effect of gender and defensive opponent on the biomechanics of sidestep cutting. *Med Sci Sports Exerc, 36*(6), 1008-1016.

McLean, S. G., Neal, R. J., Myers, P. T., & Walters, M. R. (1999). Knee joint kinematics during the sidestep cutting maneuver: potiential for injury in women. *Med Sci Sports Exerc, 31*(7), 959-968.

McLean, S. G., & Samorezov, J. E. (2009). Fatigue-induced ACL injury risk stems from a degradation in central control. *Med Sci Sports Exerc, 41*(8), 1661-1672.

McMurray, R. G., Harrell, J. S., Bangdiwala, S. I., & Hu, J. (2003). Tracking of physical activity and aerobic power from childhood through adolescence. *Med Sci Sports Exerc, 35*(11), 1914-1922.

McNitt-Gray, J. L. (1993). Kinetics of the lower extremities during drop landings from three heights. *J Biomech, 26*(9), 1037-1046.

Meyer, E. G., & Haut, R. C. (2008). Anterior cruciate ligament injury induced by internal tibial torsion or tibiofemoral compression. *J Biomech, 41*(16), 3377-3383.

Milliken, L. A., Faigenbaum, A. D., Loud, R. L., & Westcott, W. L. (2008). Correlates of upper and lower body muscular strength in children. *J Strength Cond Res, 22*(4), 1339-1346.

Montgomery, M. M., Shultz, S. J., Schmitz, R. J., Wideman, L., & Henson, R. A. (2012). Influence of Lean Body Mass and Strength on Landing Energetics. *Med Sci Sports Exerc*.

More, R. C., Karras, B. T., Neiman, R., Fritschy, D., Woo, S. L., & Daniel, D. M. (1993). Hamstrings--an anterior cruciate ligament protagonist. An in vitro study. *Am J Sports Med, 21*(2), 231-237.

Morris, N. M., & Udry, J. R. (1980). Validation of a self-administered instrument to assess state of adolescent development. *J Youth Adolesc, 9*(3), 271-280.

Myer, G. D., Ford, K. R., Barber Foss, K. D., Liu, C., Nick, T. G., & Hewett, T. E. (2009). The relationship of hamstrings and quadriceps strength to anterior cruciate ligament injury in female athletes. *Clin J Sport Med, 19*(1), 3-8.

Myer, G. D., Ford, K. R., Brent, J. L., & Hewett, T. E. (2012). An Integrated Approach to Change the Outcome Part I: Neuromuscular Screening Methods to Identify High ACL Injury Risk Athletes. *J Strength Cond Res, 26*(8), 2265-2271.

Myer, G. D., Ford, K. R., & Hewett, T. E. (2011). New method to identify athletes at high risk of ACL injury using clinic-based measurements and freeware computer analysis. *Br J Sports Med, 45*(4), 238-244.

Myer, G. D., Ford, K. R., Khoury, J., Succop, P., & Hewett, T. E. (2010). Development and validation of a clinic-based prediction tool to identify female athletes at high risk for anterior cruciate ligament injury. *Am J Sports Med, 38*(10), 2025-2033.

Nagano, A., Komura, T., & Fukashiro, S. (2007). Optimal coordination of maximal-effort horizontal and vertical jump motions--a computer simulation study. *Biomed Eng Online, 6*, 20.

Nawata, K., Teshima, R., Morio, Y., Hagino, H., Enokida, M., & Yamamoto, K. (1999). Anterior-posterior knee laxity increased by exercise. Quantitative evaluation of physiologic changes. *Acta Orthop Scand, 70*(3), 261-264.

Nguyen, A. D., & Shultz, S. J. (2007). Sex differences in clinical measures of lower extremity alignment. *J Orthop Sports Phys Ther, 37*(7), 389-398.

Oerter, K. E., Uriarte, M. M., Rose, S. R., Barnes, K. M., & Cutler, G. B., Jr. (1990). Gonadotropin secretory dynamics during puberty in normal girls and boys. *J Clin Endocrinol Metab, 71*(5), 1251-1258.

Oh, Y. K., Lipps, D. B., Ashton-Miller, J. A., & Wojtys, E. M. (2012). What strains the anterior cruciate ligament during a pivot landing? *Am J Sports Med, 40*(3), 574-583.

Olsen, O. E., Myklebust, G., Engebretsen, L., & Bahr, R. (2004). Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. *Am J Sports Med, 32*(4), 1002-1012.

Ortega, F. B., Artero, E. G., Ruiz, J. R., Vicente-Rodriguez, G., Bergman, P., Hagstromer, M., et al. (2008). Reliability of health-related physical fitness tests in European adolescents. The HELENA Study. *Int J Obes (Lond), 32 Suppl 5*, S49-57.

Padua, D. A., Marshall, S. W., Boling, M. C., Thigpen, C. A., Garrett, W. E., Jr., & Beutler, A. I. (2009). The Landing Error Scoring System (LESS) Is a valid and reliable clinical assessment tool of jump-landing biomechanics: The JUMP-ACL study. *Am J Sports Med, 37*(10), 1996-2002.

Pandy, M. G., & Shelburne, K. B. (1997). Dependence of cruciate-ligament loading on muscle forces and external load. *J Biomech, 30*(10), 1015-1024.

Parkkari, J., Pasanen, K., Mattila, V. M., Kannus, P., & Rimpela, A. (2008). The risk for a cruciate ligament injury of the knee in adolescents and young adults: a population-based cohort study of 46 500 people with a 9 year follow-up. *Br J Sports Med, 42*(6), 422-426.

Peterson, L., Junge, A., Chomiak, J., Graf-Baumann, T., & Dvorak, J. (2000). Incidence of football injuries and complaints in different age groups and skill-level groups. *Am J Sports Med, 28*(5 Suppl), S51-57.

Pollard, C. D., Sigward, S. M., Ota, S., Langford, K., & Powers, C. M. (2006). The influence of in-season injury prevention training on lower-extremity kinematics during landing in female soccer players. *Clin J Sport Med, 16*(3), 223-227.

Pollard, C. D., Sigward, S. M., & Powers, C. M. (2010). Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. *Clin Biomech (Bristol, Avon), 25*(2), 142-146.

Price, R. J., Hawkins, R. D., Hulse, M. A., & Hodson, A. (2004). The Football Association medical research programme: an audit of injuries in academy youth football. *Br J Sports Med, 38*(4), 466-471.

Quammen, D., Cortes, N., Van Lunen, B. L., Lucci, S., Ringleb, S. I., & Onate, J. (2012). Two different fatigue protocols and lower extremity motion patterns during a stop-jump task. *J Athl Train, 47*(1), 32-41.

Quatman, C. E., Ford, K. R., Myer, G. D., & Hewett, T. E. (2006). Maturation leads to gender differences in landing force and vertical jump performance: a longitudinal study. *Am J Sports Med, 34*(5), 806-813.

Quatman, C. E., Ford, K. R., Myer, G. D., Paterno, M. V., & Hewett, T. E. (2008). The effects of gender and pubertal status on generalized joint laxity in young athletes. *J Sci Med Sport, 11*(3), 257-263.

Ramsbottom, R., Brewer, J., & Williams, C. (1988). A progressive shuttle run test to estimate maximal oxygen uptake. *Br J Sports Med, 22*(4), 141-144.

Rapkin, A. J., Tsao, J. C., Turk, N., Anderson, M., & Zeltzer, L. K. (2006). Relationships among self-rated tanner staging, hormones, and psychosocial factors in healthy female adolescents. *J Pediatr Adolesc Gynecol, 19*(3), 181-187.

Renstrom, P., Ljungqvist, A., Arendt, E., Beynnon, B., Fukubayashi, T., Garrett, W., et al. (2008). Non-contact ACL injuries in female athletes: an International Olympic Committee current concepts statement. *Br J Sports Med, 42*(6), 394-412.

Rodriguez, G., Moreno, L. A., Blay, M. G., Blay, V. A., Fleta, J., Sarria, A., et al. (2005). Body fat measurement in adolescents: comparison of skinfold thickness equations with dual-energy X-ray absorptiometry. *Eur J Clin Nutr, 59*(10), 1158-1166.

Roewer, B. D., Di Stasi, S. L., & Snyder-Mackler, L. (2011). Quadriceps strength and weight acceptance strategies continue to improve two years after anterior cruciate ligament reconstruction. *J Biomech, 44*(10), 1948-1953.

Rogol, A. D., Clark, P. A., & Roemmich, J. N. (2000). Growth and pubertal development in children and adolescents: effects of diet and physical activity. *Am J Clin Nutr, 72*(2 Suppl), 521S-528S.

Rose, S. R., Municchi, G., Barnes, K. M., Kamp, G. A., Uriarte, M. M., Ross, J. L., et al. (1991). Spontaneous growth hormone secretion increases during puberty in normal girls and boys. *J Clin Endocrinol Metab, 73*(2), 428-435.

Rudolph, K. S., Axe, M. J., Buchanan, T. S., Scholz, J. P., & Snyder-Mackler, L. (2001). Dynamic stability in the anterior cruciate ligament deficient knee. *Knee Surg Sports Traumatol Arthrosc, 9*(2), 62-71.

Ruiz, J. R., Castro-Pinero, J., Espana-Romero, V., Artero, E. G., Ortega, F. B., Cuenca, M. M., et al. (2011). Field-based fitness assessment in young people: the ALPHA health-related fitness test battery for children and adolescents. *Br J Sports Med, 45*(6), 518-524.

Ruiz, J. R., Ramirez-Lechuga, J., Ortega, F. B., Castro-Pinero, J., Benitez, J. M., Arauzo-Azofra, A., et al. (2008). Artificial neural network-based equation for estimating VO2max from the 20 m shuttle run test in adolescents. *Artif Intell Med, 44*(3), 233-245.

Ruiz, J. R., Silva, G., Oliveira, N., Ribeiro, J. C., Oliveira, J. F., & Mota, J. (2009). Criterion-related validity of the 20-m shuttle run test in youths aged 13-19 years. *J Sports Sci, 27*(9), 899-906.

Rutenfranz, J., Macek, M., Lange Andersen, K., Bell, R. D., Vavra, J., Radvansky, J., et al. (1990). The relationship between changing body height and growth related changes in maximal aerobic power. *Eur J Appl Physiol Occup Physiol, 60*(4), 282-287.

Sakane, M., Livesay, G. A., Fox, R. J., Rudy, T. W., Runco, T. J., & Woo, S. L. (1999). Relative contribution of the ACL, MCL, and bony contact to the anterior stability of the knee. *Knee Surg Sports Traumatol Arthrosc, 7*(2), 93-97.

Sanna, G., & O'Connor, K. M. (2008). Fatigue-related changes in stance leg mechanics during sidestep cutting maneuvers. *Clin Biomech (Bristol, Avon), 23*(7), 946-954.

Schmitz, K. E., Hovell, M. F., Nichols, J. F., Irvin, V. L., Keating, K., Simon, G. M., et al. (2004). A validation study of early adolescents' pubertal self-assessments. *Journal of Early Adolescence, 24*(4), 357-384.

Schmitz, R. J., & Shultz, S. J. (2010). Contribution of knee flexor and extensor strength on sex-specific energy absorption and torsional joint stiffness during drop jumping. *J Athl Train, 45*(5), 445-452.

Schmitz, R. J., Shultz, S. J., & Nguyen, A. D. (2009). Dynamic valgus alignment and functional strength in males and females during maturation. *J Athl Train, 44*(1), 26-32.

Shea, K. G., Pfeiffer, R., Wang, J. H., Curtin, M., & Apel, P. J. (2004). Anterior cruciate ligament injury in pediatric and adolescent soccer players: an analysis of insurance data. *J Pediatr Orthop, 24*(6), 623-628.

Shultz, S. J., Nguyen, A. D., Leonard, M. D., & Schmitz, R. J. (2009). Thigh strength and activation as predictors of knee biomechanics during a drop jump task. *Med Sci Sports Exerc, 41*(4), 857-866.

Shultz, S. J., Nguyen, A. D., & Schmitz, R. J. (2008). Differences in lower extremity anatomical and postural characteristics in males and females between maturation groups. *J Orthop Sports Phys Ther, 38*(3), 137-149.

Shultz, S. J., & Schmitz, R. J. (2009). Effects of transverse and frontal plane knee laxity on hip and knee neuromechanics during drop landings. *Am J Sports Med, 37*(9), 1821-1830.

Shultz, S. J., Schmitz, R. J., Cone, J. R., Copple, T. J., Montgomery, M. M., Pye, M. L., et al. (2013). Multiplanar Knee Laxity Increases during a 90-min Intermittent Exercise Protocol. *Med Sci Sports Exerc*.

Shultz, S. J., Schmitz, R. J., Cone, J. R., Henson, R., Montgomery, M. M., Pye, M. L., et al. (2013). Knee Laxity Increases during Intermittent Exercise Influence Landing Biomechanics. *Med Sci Sports Exerc, In Review*.

Shultz, S. J., Schmitz, R. J., Kong, Y., Dudley, W. N., Beynnon, B. D., Nguyen, A. D., et al. (2012). Cyclic variations in multiplanar knee laxity influence landing biomechanics. *Med Sci Sports Exerc, 44*(5), 900-909.

Shultz, S. J., Schmitz, R. J., Nguyen, A. D., Chaudhari, A. M., Padua, D. A., McLean, S. G., et al. (2010). ACL Research Retreat V: an update on ACL injury risk and prevention, March 25-27, 2010, Greensboro, NC. *J Athl Train, 45*(5), 499-508.

Shultz, S. J., Schmitz, R. J., Nguyen, A. D., & Levine, B. J. (2010). Joint laxity is related to lower extremity energetics during a drop jump landing. *Med Sci Sports Exerc, 42*(4), 771-780.

Shultz, S. J., Schmitz, R. J., Tritsch, A. J., & Montgomery, M. M. (2012). Methodological considerations of task and shoe wear on joint energetics during landing. *J Electromyogr Kinesiol, 22*(1), 124-130.

Sigward, S. M., Pollard, C. D., & Powers, C. M. (2008). The influence of an ACL injury prevention program on knee valgus moments during cutting; an evaluation of pre-pubertal, pubertal and post-pubertal female athletes. *J Athl Train, 43*(5), 559.

Sigward, S. M., Pollard, C. D., & Powers, C. M. (2011). The influence of sex and maturation on landing biomechanics: implications for anterior cruciate ligament injury. *Scand J Med Sci Sports*.

Sigward, S. M., & Powers, C. M. (2006). The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. *Clin Biomech (Bristol, Avon), 21*(1), 41-48.

Slaughter, M. H., Lohman, T. G., Boileau, R. A., Horswill, C. A., Stillman, R. J., Van Loan, M. D., et al. (1988). Skinfold equations for estimation of body fatness in children and youth. *Hum Biol, 60*(5), 709-723.

Smith, H. C., Johnson, R. J., Shultz, S. J., Tourville, T., Holterman, L. A., Slauterbeck, J., et al. (2012). A prospective evaluation of the Landing Error Scoring System (LESS) as a screening tool for anterior cruciate ligament injury risk. *Am J Sports Med, 40*(3), 521-526.

Souza, R. B., Fang, C., Luke, A., Wu, S., Li, X., & Majumdar, S. (2012). Relationship between knee kinetics during jumping tasks and knee articular cartilage MRI T1rho and T2 relaxation times. *Clin Biomech (Bristol, Avon), 27*(4), 403-408.

Steinberger, J., Jacobs, D. R., Raatz, S., Moran, A., Hong, C. P., & Sinaiko, A. R. (2005). Comparison of body fatness measurements by BMI and skinfolds vs dual energy X-ray absorptiometry and their relation to cardiovascular risk factors in adolescents. *Int J Obes (Lond), 29*(11), 1346-1352.

Swartz, E. E., Decoster, L. C., Russell, P. J., & Croce, R. V. (2005). Effects of Developmental Stage and Sex on Lower Extremity Kinematics and Vertical Ground Reaction Forces During Landing. *J Athl Train, 40*(1), 9-14.

Taeymans, J., Clarys, P., Abidi, H., Hebbelinck, M., & Duquet, W. (2009). Developmental changes and predictability of static strength in individuals of different maturity: a 30-year longitudinal study. *J Sports Sci, 27*(8), 833-841.

Tanner, J. M. (1962). *Growth at adolescence; with a general consideration of the effects of hereditary and environmental factors upon growth and maturation from birth to maturity* (2d ed.). Springfield, Ill.

Thomas, A. C., McLean, S. G., & Palmieri-Smith, R. M. (2010). Quadriceps and hamstrings fatigue alters hip and knee mechanics. *J Appl Biomech, 26*(2), 159-170.

Toyoda, T., Matsumoto, H., Fujikawa, K., Saito, S., & Inoue, K. (1998). Tensile load and the metabolism of anterior cruciate ligament cells. *Clin Orthop Relat Res*(353), 247-255.

Treuth, M. S., Butte, N. F., Wong, W. W., & Ellis, K. J. (2001). Body composition in prepubertal girls: comparison of six methods. *Int J Obes Relat Metab Disord, 25*(9), 1352-1359.

Uhorchak, J. M., Scoville, C. R., Williams, G. N., Arciero, R. A., St Pierre, P., & Taylor, D. C. (2003). Risk factors associated with noncontact injury of the anterior cruciate ligament: a prospective four-year evaluation of 859 West Point cadets. *Am J Sports Med, 31*(6), 831-842.

Walsh, M. S., Waters, J., & Kersting, U. G. (2007). Gender bias on the effects of instruction on kinematic and kinetic jump parameters of high-level athletes. *Res Sports Med, 15*(4), 283-295.

Wild, C. Y., Steele, J. R., & Munro, B. J. (2012). Why do girls sustain more anterior cruciate ligament injuries than boys?: a review of the changes in estrogen and musculoskeletal structure and function during puberty. *Sports Med, 42*(9), 733-749.

Wild, C. Y., Steele, J. R., & Munro, B. J. (2013a). Insufficient hamstring strength compromises landing technique in adolescent girls. *Med Sci Sports Exerc, 45*(3), 497-505.

Wild, C. Y., Steele, J. R., & Munro, B. J. (2013b). Musculoskeletal and estrogen changes during the adolescent growth spurt in girls. *Med Sci Sports Exerc, 45*(1), 138-145.

Winter, D. A. (2009). *Biomechanics and Motor Control of Human Movement* (4th ed.). Hoboken, NJ: John Wiley & Sons, Inc.

Withrow, T. J., Huston, L. J., Wojtys, E. M., & Ashton-Miller, J. A. (2006). The effect of an impulsive knee valgus moment on in vitro relative ACL strain during a simulated jump landing. *Clin Biomech (Bristol, Avon), 21*(9), 977-983.

Woo, S. L. Y., Fox, R. J., Sakane, M., Livesay, G. A., Rudy, T. W., & Fu, F. H. (1998). Biomechanics of the ACL: Measurements of in situ force in the ACL and knee kinematics. *Knee, 5*(4), 267-288.

Woodhouse, E., Schmale, G. A., Simonian, P., Tencer, A., Huber, P., & Seidel, K. (2007). Reproductive hormone effects on strength of the rat anterior cruciate ligament. *Knee Surg Sports Traumatol Arthrosc, 15*(4), 453-460.

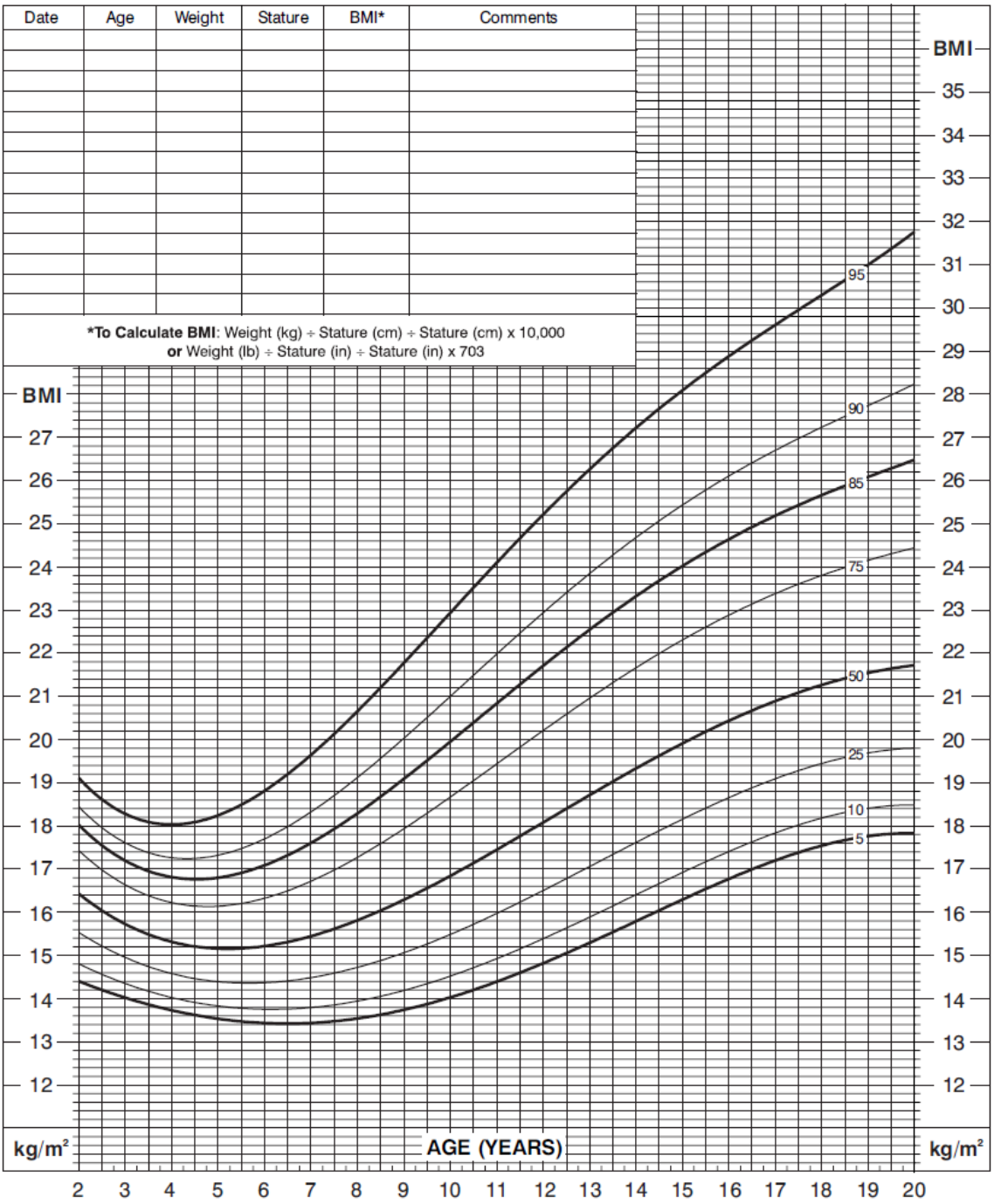
Yasuda, K., & Sasaki, T. (1987). Exercise after anterior cruciate ligament reconstruction. The force exerted on the tibia by the separate isometric contractions of the quadriceps or the hamstrings. *Clin Orthop Relat Res*(220), 275-283.

Yu, B., McClure, S. B., Onate, J. A., Guskiewicz, K. M., Kirkendall, D. T., & Garrett, W. E. (2005). Age and gender effects on lower extremity kinematics of youth soccer players in a stop-jump task. *Am J Sports Med, 33*(9), 1356-1364.

Zhang, S., Bates, B. T., & Dufek, J. S. (2000). Contributions of lower extremity joints to energy dissipation during landings. *Med Sci Sports Exerc, 32*(4), 812-819.

APPENDIX A

BODY MASS INDEX FOR AGE PERCENTILES IN FEMALES AGE 2 TO 20



APPENDIX B

INSTITUTIONAL REVIEW BOARD CONSENT FORM

UNIVERSITY OF NORTH CAROLINA AT GREENSBORO
CONSENT FOR A MINOR TO ACT AS A HUMAN PARTICIPANT: LONG FORM

Project Title: Associations Between Physical Characteristics and Landing Mechanics in Adolescent Females

Project Director: Amanda J Tritsch, Sandra J Shultz

UNCG IRB
Approved Consent Form
Valid 8/28/12 to 8/27/13

Participant's Name: _____

What is the study about?
This is a research project. The purpose of the study is to evaluate the measurement consistency of modifiable physical assessment factors and a jumping and landing task.

Why are you asking my child?
You child is being asked to participate because they are between the ages of 12 and 15 years of age, have no current history of injury to the lower extremity, or any current pain or injury to the lower extremity which may affect their ability to perform jumping activities.

What will you ask my child to do if I agree to let him or her be in the study?
Your child will be asked to attend a 1.5 hour testing session. Prior to any testing, you and your child will complete a health history and activity questionnaire, and a physical activity readiness questionnaire (PAR-Q) to screen for any conditions that might exclude your child from the study. They will also be asked to complete a self-reported Tanner stage of maturation questionnaire, will be completed in a private office. During the first testing session an investigator will record your child's height, weight, and age, and take measures of your child's waist, hip, and arm using standard measurement devices (ruler, goniometer, anthropometric calipers, etc). During this time, we will also measure your child's grip strength, jumping ability, and single leg reaching ability. Then we'll teach them how to perform movement tasks which include a drop jump task and hopping movements which resemble those performed during sport activities. The drop jump task will be performed from 45cm and require them to drop off of a box and then perform a maximal vertical jump. Your child will perform a fitness test and 5 of the drop jumps. The drop jumps will be performed both before and after the fitness test. This testing will take place in the Applied Neuromechanics Research Laboratory in HHP 238 on the campus of UNCG.

What are the dangers to my child?
There is minimal risk of joint strain or muscle pull during the drop jump task (will be performed during both sessions) and fitness test (which will only be performed during the second session)

The only test that may cause some discomfort is the skin fold measurement, which requires the investigator to grasp the skin on the front and rear of the arm between their forefinger and thumb. Thus, your child may feel a small pinch during this test.

To minimize the risk of any discomfort associated with the self-reported Tanner stage of maturation questionnaire, your child will fill out the forms in a private office. The questionnaire will be given to your child in a plain manila envelope and they will be instructed to place the questionnaire back in the envelope and leave it in the office upon completion. If they have any questions while filling out the questionnaire, the primary researcher will be available to answer them.

The Institutional Review Board at the University of North Carolina at Greensboro has determined that participation in this study poses minimal risk to participants.

If you have any concerns about your child rights, how she is being treated or if you have questions, want more information or have suggestions, please contact Eric Allen in the Office of Research Compliance at UNCG at (336)

256-1482. Questions, concerns or complaints about this project or benefits or risks associated with being in this study can be answered by Dr. Sandra J. Shultz who may be contacted at (336) 334-3027 (sjshultz@uncg.edu).

Are there any benefits to my child as a result of participation in this research study?
There are no direct benefits to you for participating in this study.

Are there any benefits to society as a result of my child taking part in this research?
There are no direct benefits to society as a result of this study, though results from this study may help us to better understand how physical characteristics that change with maturation affect movement patterns and risk of knee injury.

Will my child get paid for being in the study? Will it cost me anything for my kid to be in this study?
Your child will receive a $25 gift card upon completion of the study. There are no costs to you for participating in this study.

How will my child's information be kept confidential?
All information obtained in this study is strictly confidential unless disclosure is required by law. All consent forms will be maintained in a locked file only accessible by the investigator for three years, at which time they will be destroyed by shredding. Your information and data will be assigned a code number. The list connecting your child's name to this number will be kept in a locked file until the study has been completed and data analyzed, at which time the list will be destroyed. Your child's name will not be used in any report. All data will be stored on the principal investigators personal computer identified only by subject number. These data will be kept indefinitely. A photocopy of this original consent form will be provided to you for your records.

What if my child wants to leave the study or I want him/her to leave the study?
You have the right to refuse to allow your child to participate or to withdraw him or her at any time, without penalty. If your child does withdraw, it will not affect you or your child in any way. If you or your child chooses to withdraw, you may request that any data which has been collected be destroyed unless it is in a de-identifiable state.

What about new information/changes in the study?
If significant new information relating to the study becomes available which may relate to your willingness allow your child to continue to participate, this information will be provided to you.

Voluntary Consent by Participant:
By signing this consent form, you are agreeing that you have read it or it has been read to you. You fully understand the contents of this document and consent to your child taking part in this study. All of your questions concerning this study have been answered. By signing this form, you are agreeing that you are the legal parent or guardian of the child who wishes to participate in this study described to you by Amanda Tritsch.

_____ Date: _____
Participant's Parent/Legal Guardian's Signature

UNCG IRB
Approved Consent Form
Valid 8/28/12 to 8/27/13

APPENDIX C

INJURY HISTORY

PHYSICAL ACTIVITY AND HEALTH HISTORY

Do you have any General Health Problems or Illnesses? (e.g. diabetes, respiratory disease) Yes____ No____

Do you have any vestibular (inner ear) or balance disorders? Yes____ No____

Do you smoke? Yes____ No____

Do you drink alcohol? Yes____ No____ If yes, how often?_____

Do you have any history of connective tissue disease or disorders? (e.g. Ehlers-Danlos, Marfan's Syndrome, Rheumatoid Arthritis) Yes____ No____

Has a family member of yours ever been diagnosed with breast cancer? Yes____ No____ (if no, please skip next question.)

If yes, please put a check next to the types of relatives that have been diagnosed. You may check more than one box:

Mother_____ Sister_____ Grandmother _____ Aunt _____.
Male relative (father, brother, grandfather, or uncle) _____.
Other type of relative (please write in) _____.

Please list any medications you take regularly: _____
_____

Please list any previous injuries to your lower extremities. Please include a description of the injury (e.g. ligament sprain, muscle strain), severity of the injury, date of the injury, and whether it was on the left or right side.

Body Part	Description	Severity	Date of Injury	L or R
Hip				
Thigh				
Knee				

Lower Leg_____

Ankle_____

Foot_____

Please list any previous surgery to your lower extremities (Include a description of the surgery, the date of the surgery, and whether it was on the left or right side)

Body Part	Description	Date of Surgery	L or R

Please list all physical activities that you are currently engaged in. For each activity, please indicate how much time you spend each week in this activity, the intensity of the activity (i.e. competitive or recreational) and for how long you have been regularly participating in the activity.

Activity	#Days/week	#Minutes/Day	Intensity	Activity Began When?

What time of day do you generally engage in the above activities?_____

_____

Please list other conditions / concerns that you feel we should be aware of:_____

_____

_____

_____

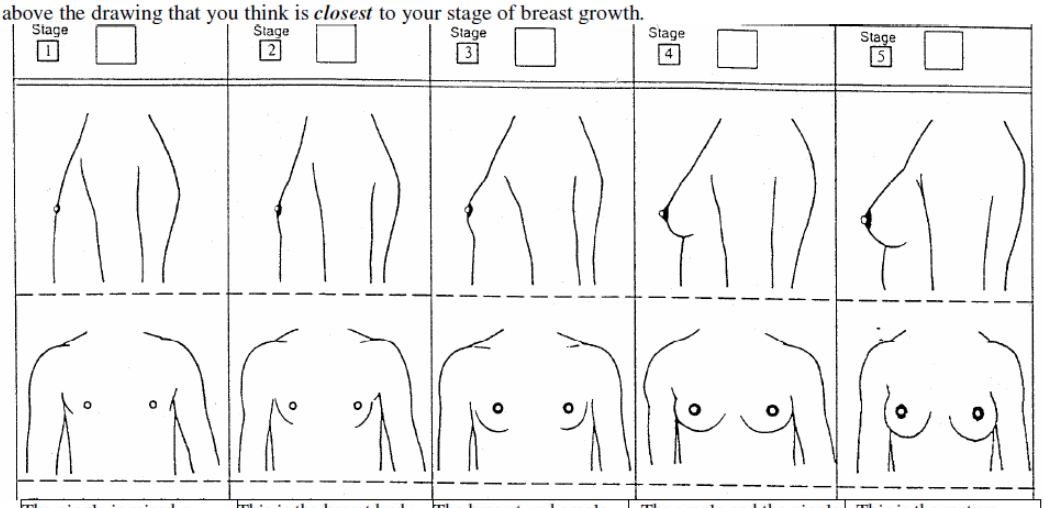
_____

APPENDIX D

TANNER STAGE OF MATURATION QUESTIONNAIRE

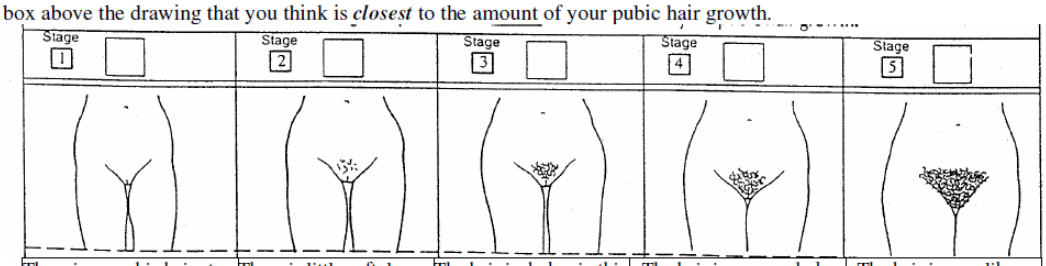
Subject #:

SET ONE: The drawings below show 5 different stages of how the breasts grow. A girl can go through each of the 5 stages as shown. Please look at each drawing and read the sentences that match the drawings. Then mark an "X" in the box above the drawing that you think is ***closest*** to your stage of breast growth.



The nipple is raised a little. The rest of the breast is still flat.	This is the breast bud stage. The nipple is raised more than in stage 1. The breast is a small mound. The areola is larger than stage 1	The breasrt and areola are both larger than in stage 2. The areola does not stick out away from the breast.	The areola and the nipple make up a mound that sticks up above the shape of the breast. NOTE: This stage may not happen at all for some girls. Some girls develop from stage 3 to stage 5 with no stage 4	This is the mature adult stage. The breasts are fully developed. Only the nipple sticks out in this stage. The areola has moved back in the general shape of the breast.

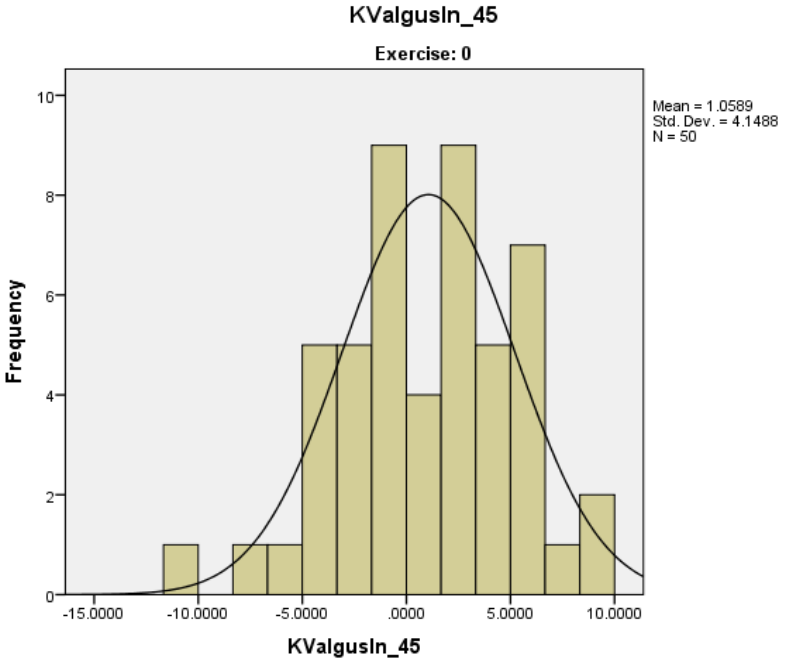
SET TWO: The drawings below show 5 different stages of female pubic hair growth.. A girl goes through each of the 5 stages as shown. Please look at each drawing and read the sentences that match the drawings. Then mark an "X" in the box above the drawing that you think is ***closest*** to the amount of your pubic hair growth.

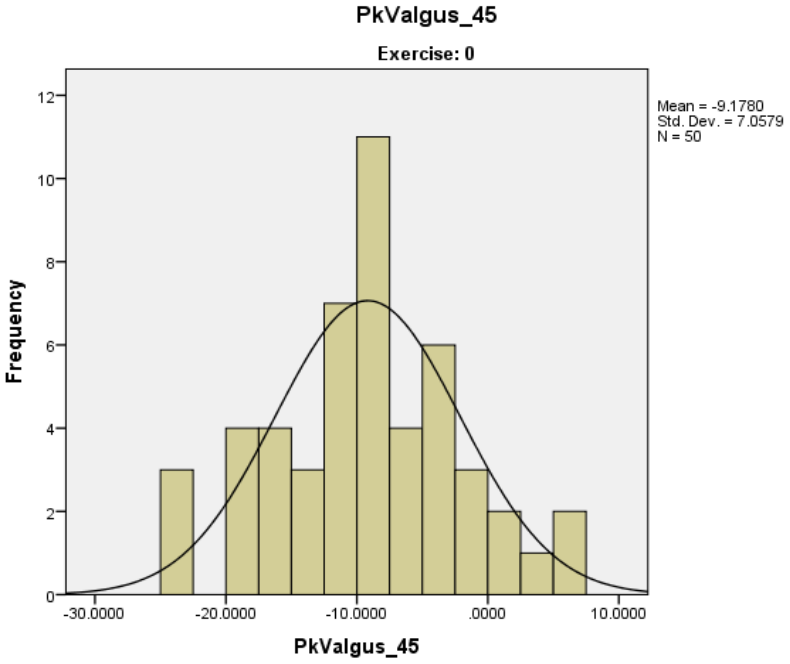


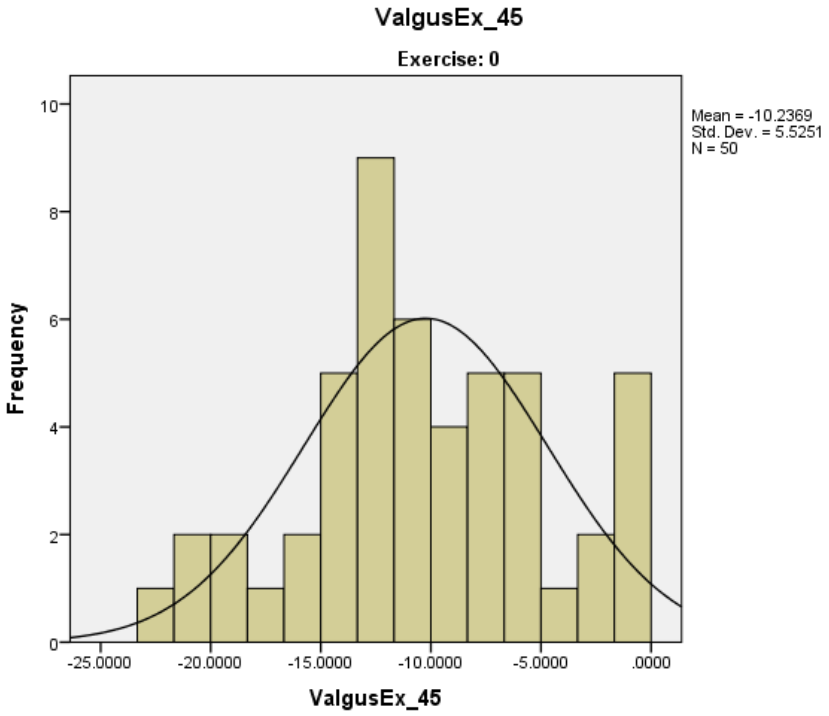
There is no pubic hair at all..	There is little soft, long lightly colored hair. This hair may be straight or a little curly.	The hair is darker in this stage. It is coarser and more curled. It has spread out and thinly covers a bigger area.	The hair is now as dark, curly, and coarse as that of an adult female. The area that the hair covers is not as big as that of an adult female. The hair has not spread out to the legs.	The hair is now like that of an adult female. It covers the same area as that of and adult female. The hair usually forms a triangular (⋁) pattern as is spreads out to the legs.

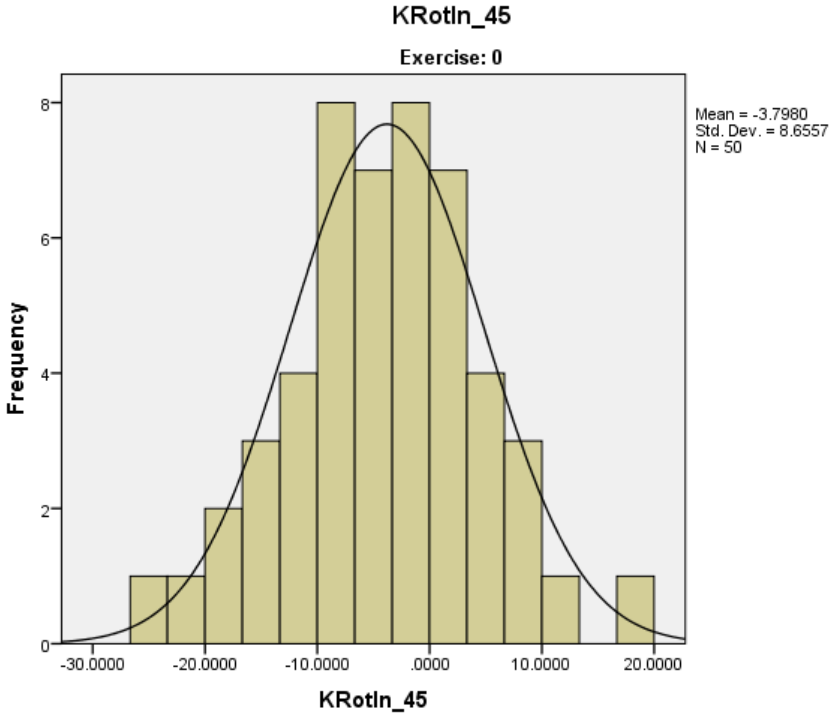
APPENDIX E

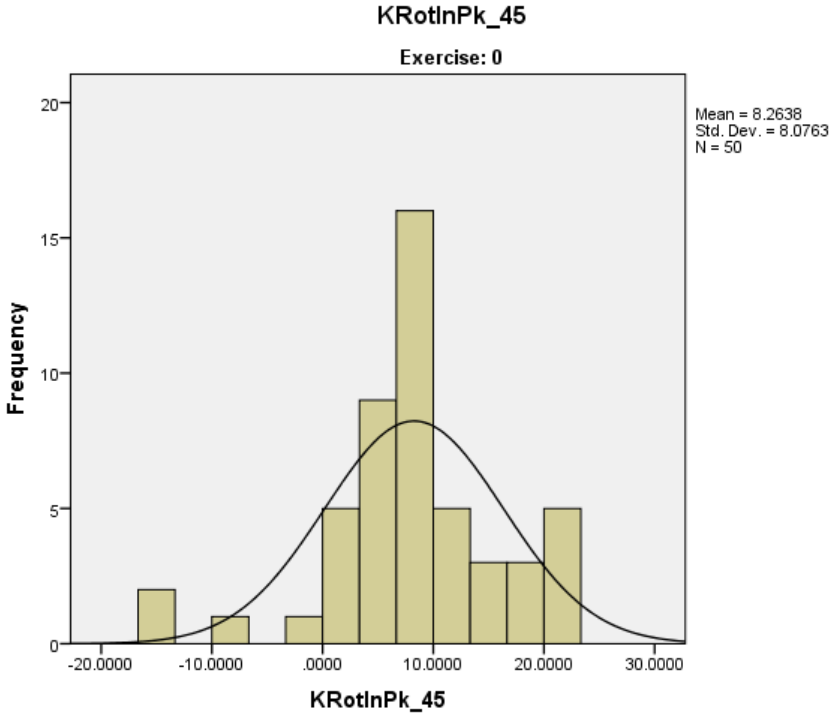
HISTOGRAMS OF DEPENDENT VARIABLES PRIOR TO EXERCISE

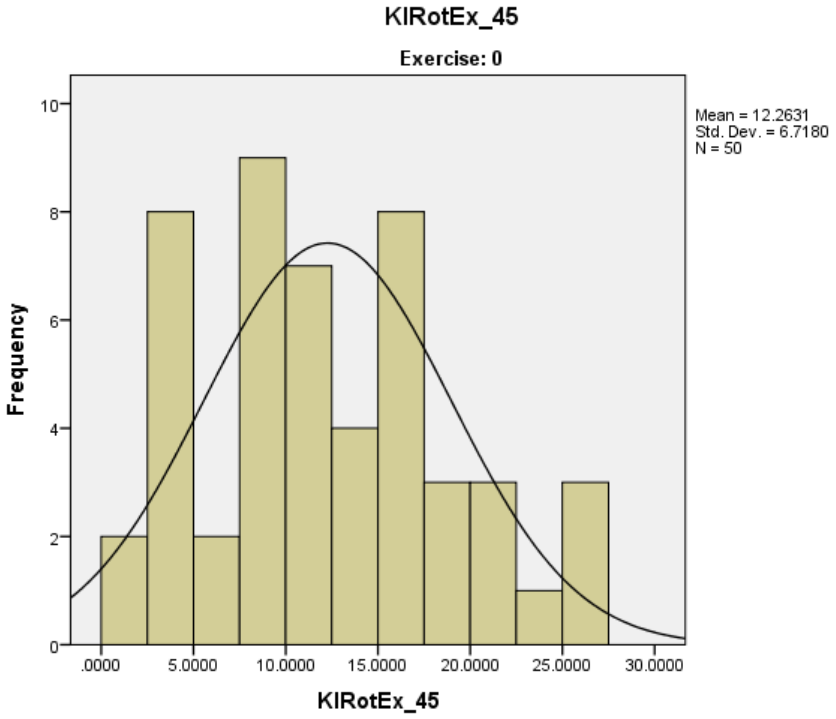


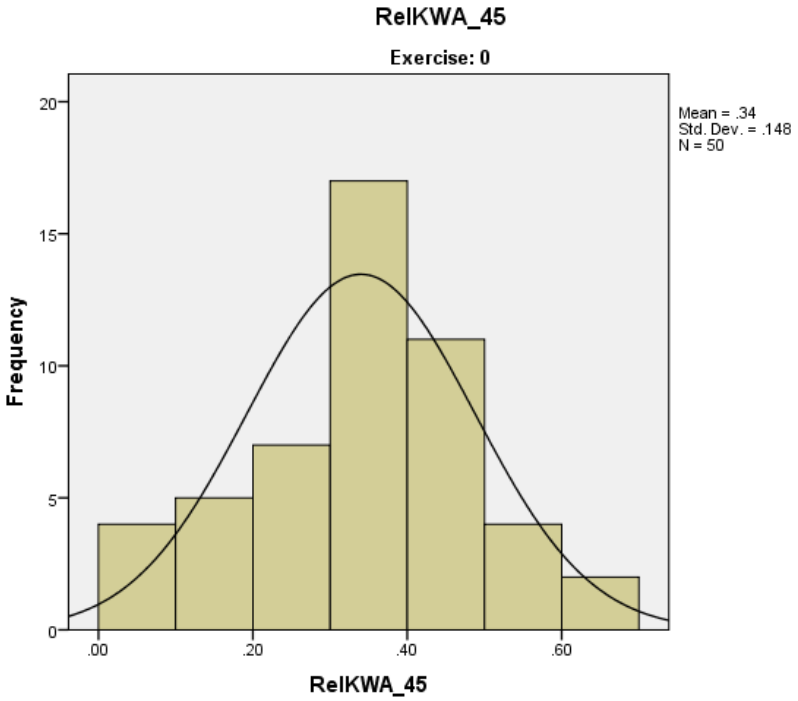



ValgusEx_45
Exercise: 0
Mean = -10.2369
Std. Dev. = 5.5251
N = 50
Frequency
0
2
4
6
8
10
-25.0000
-20.0000
-15.0000
-10.0000
-5.0000
.0000
ValgusEx_45


KRotIn_45
Exercise: 0
Mean = -3.7980
Std. Dev. = 8.6557
N = 50
Frequency
0
2
4
6
8
-30.0000
-20.0000
-10.0000
.0000
10.0000
20.0000
KRotIn_45

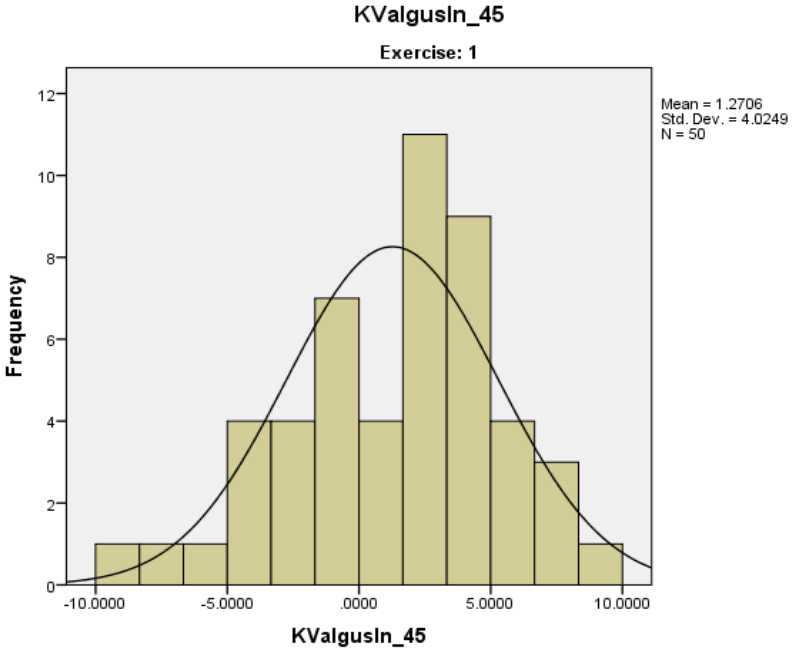

KRotInPk_45
Exercise: 0
Mean = 8.2638
Std. Dev. = 8.0763
N = 50
Frequency
20
15
10
5
0
-20.0000
-10.0000
.0000
10.0000
20.0000
30.0000
KRotInPk_45

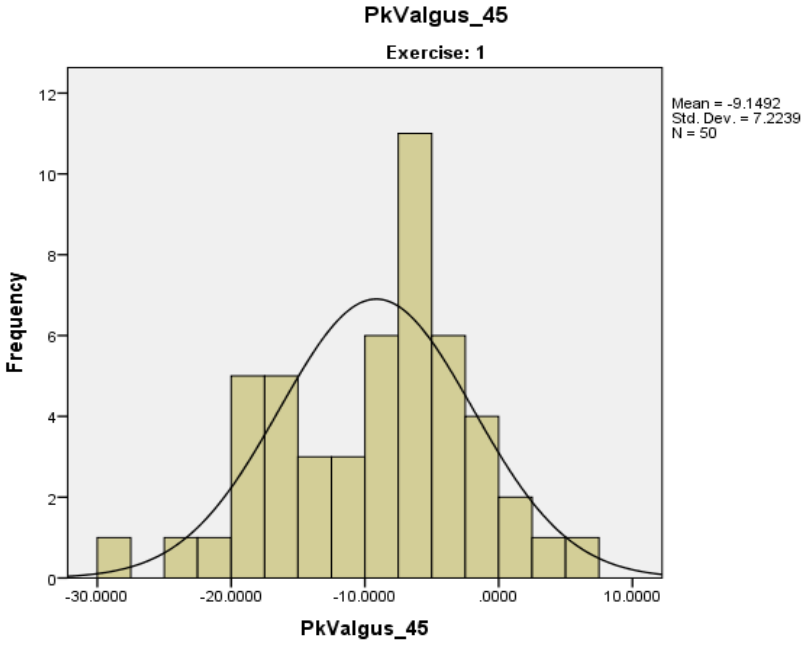

KIRotEx_45
Exercise: 0
Mean = 12.2631
Std. Dev. = 6.7180
N = 50
Frequency
10
8
6
4
2
0
.0000
5.0000
10.0000
15.0000
20.0000
25.0000
30.0000
KIRotEx_45

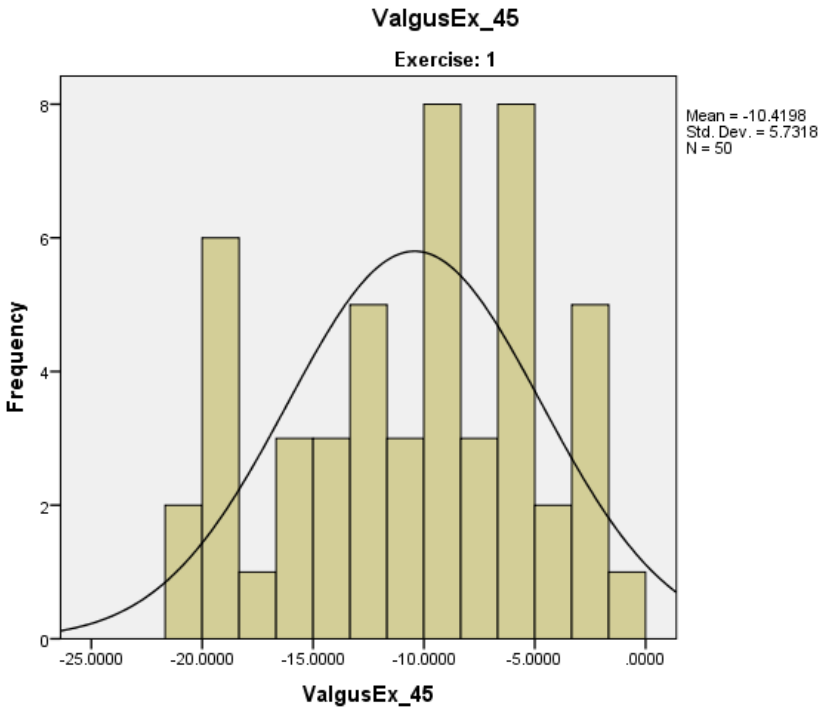

RelKWA_45
Exercise: 0
Mean = .34
Std. Dev. = .148
N = 50
Frequency
20
15
10
5
0
.00
.20
.40
.60
RelKWA_45

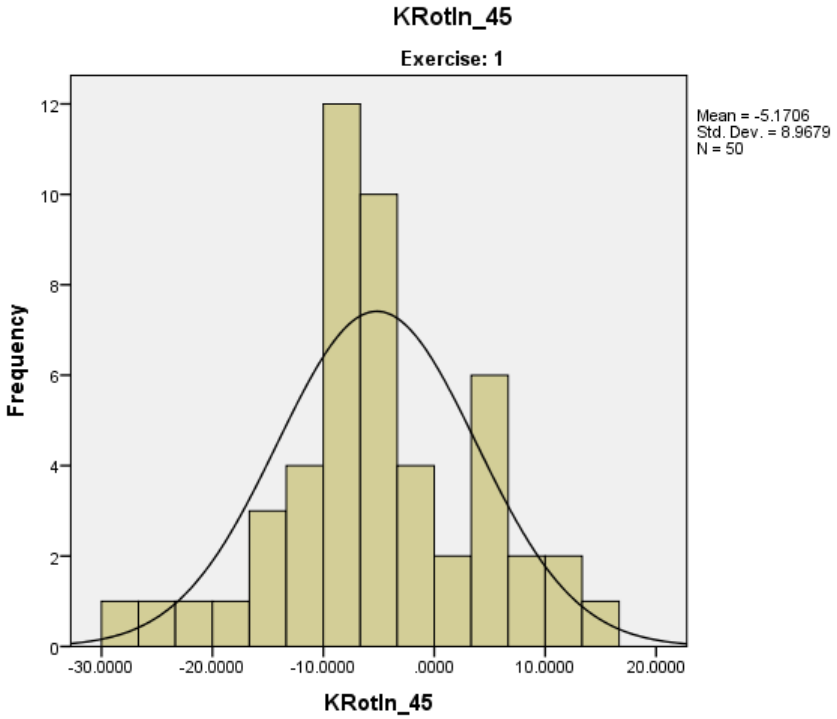
APPENDIX F

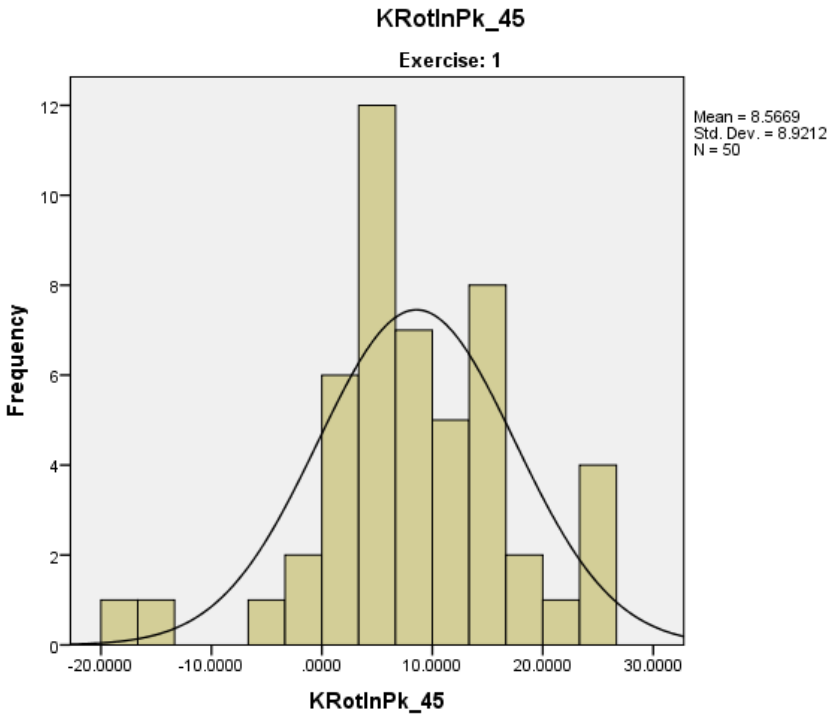
HISTOGRAMS OF DEPENDENT VARIABLES FOLLOWING EXERCISE

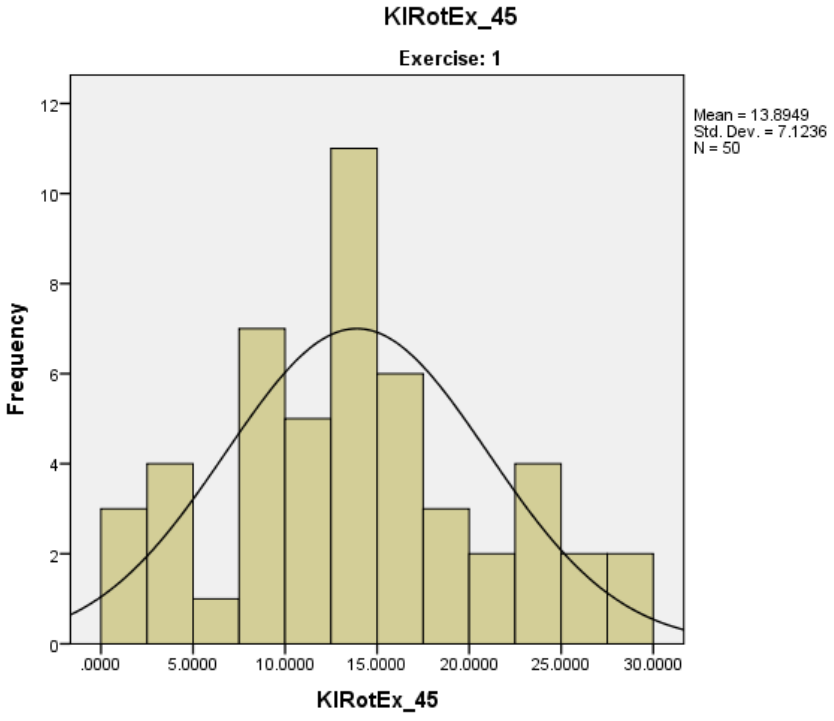


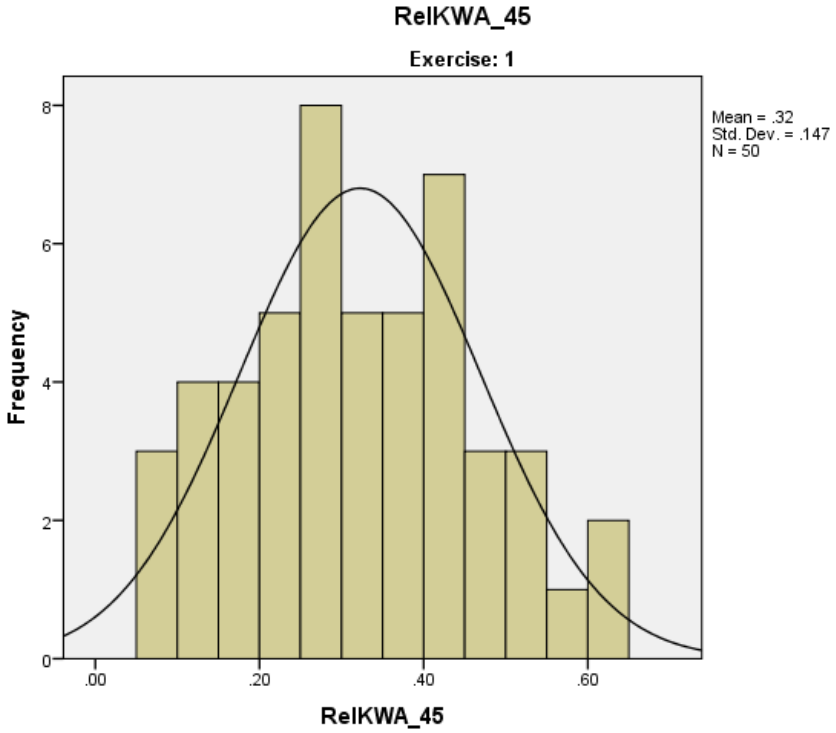



ValgusEx_45
Exercise: 1
Mean = -10.4198
Std. Dev. = 5.7318
N = 50
Frequency
0
2
4
6
8
-25.0000
-20.0000
-15.0000
-10.0000
-5.0000
.0000
ValgusEx_45


KRotIn_45
Exercise: 1
Mean = -5.1706
Std. Dev. = 8.9679
N = 50
Frequency
0
2
4
6
8
10
12
-30.0000
-20.0000
-10.0000
.0000
10.0000
20.0000
KRotIn_45

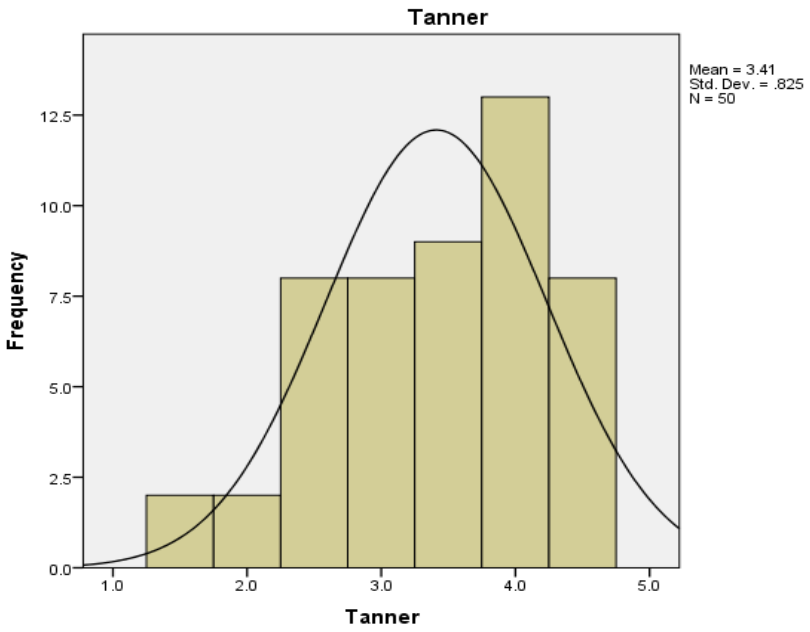

KRotInPk_45
Exercise: 1
Mean = 8.5669
Std. Dev. = 8.9212
N = 50
Frequency
0
2
4
6
8
10
12
-20.0000
-10.0000
.0000
10.0000
20.0000
30.0000
KRotInPk_45

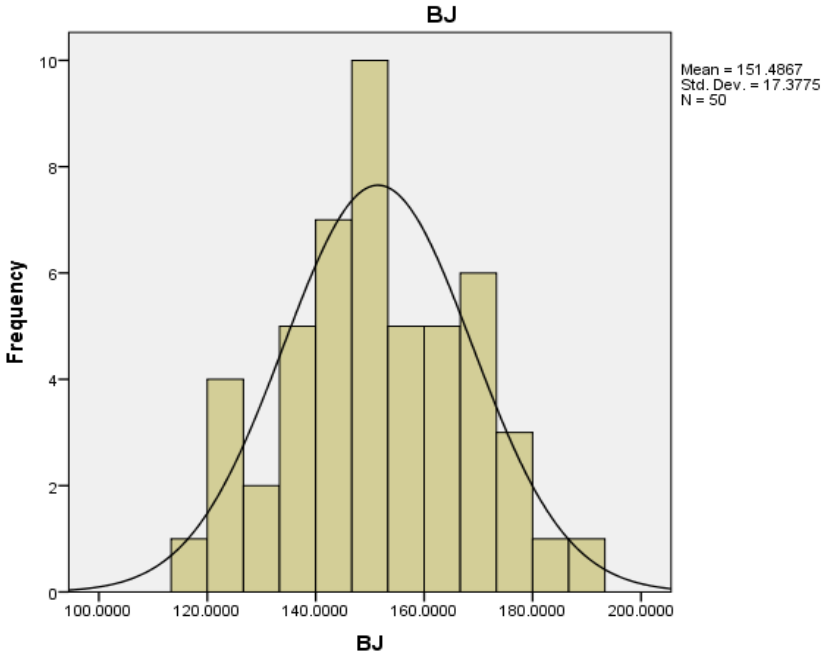

KIRotEx_45
Exercise: 1
Mean = 13.8949
Std. Dev. = 7.1236
N = 50
Frequency
0
2
4
6
8
10
12
.0000
5.0000
10.0000
15.0000
20.0000
25.0000
30.0000
KIRotEx_45

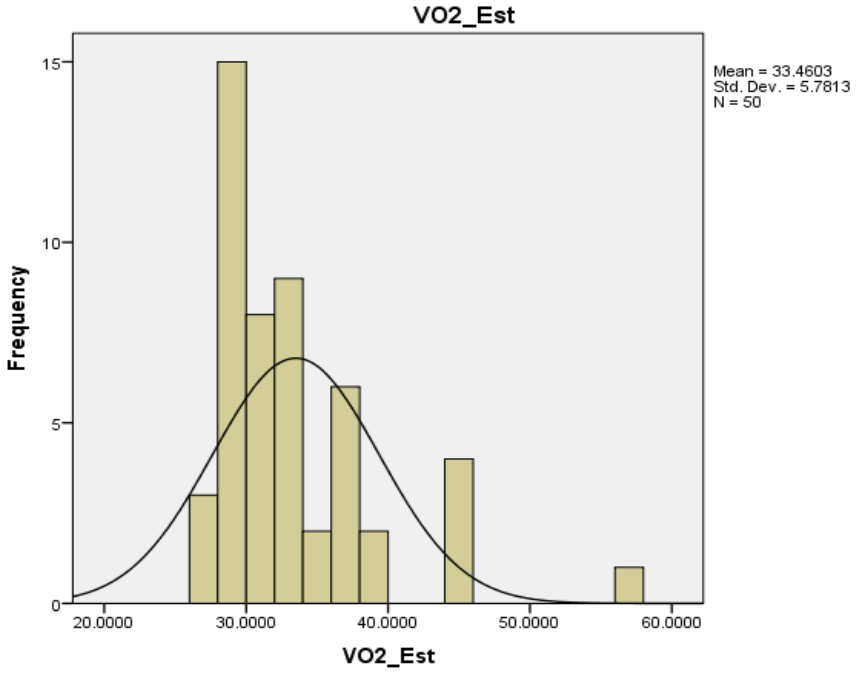

RelKWA_45
Exercise: 1
Mean = .32
Std. Dev. = .147
N = 50
Frequency
0
2
4
6
8
.00
.20
.40
.60
RelKWA_45

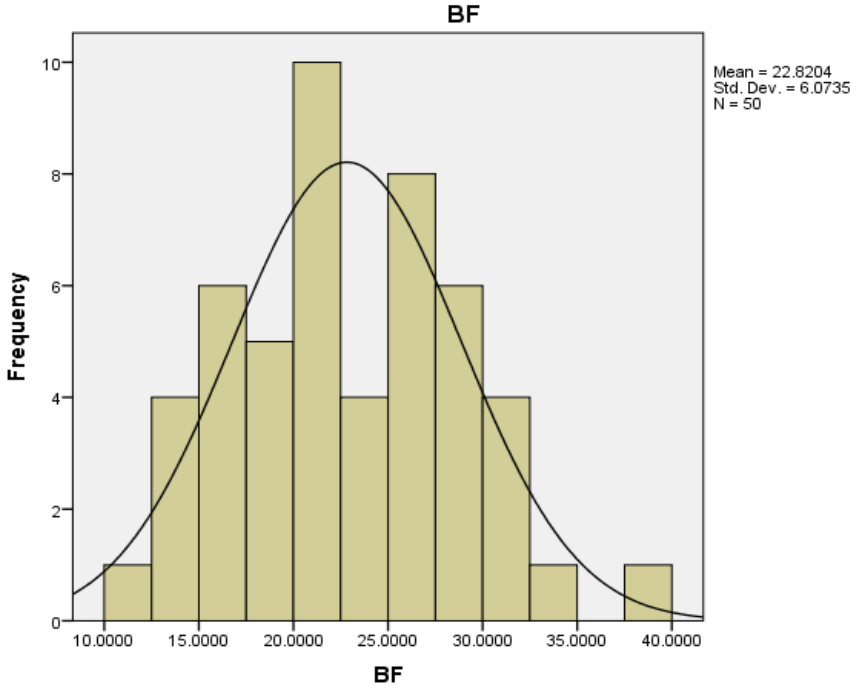
APPENDIX G

HISTOGRAMS OF PHYSICAL CHARACTERISTICS






VO2_Est
Mean = 33.4603
Std. Dev. = 5.7813
N = 50
Frequency
15
10
5
0
20.0000
30.0000
40.0000
50.0000
60.0000
VO2_Est


BF
Mean = 22.8204
Std. Dev. = 6.0735
N = 50
Frequency
10
8
6
4
2
0
10.0000
15.0000
20.0000
25.0000
30.0000
35.0000
40.0000
BF

APPENDIX H

SPSS OUTPUTS OF ALL STATISTICAL MODELS

Regression: Pre Exercise Initial Knee Valgus

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 0 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.197[a]	.039	.019	4.1099033	.039	1.933	1	48	.171
2	.262[b]	.068	-.014	4.1784292	.030	.480	3	45	.698

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	32.650	1	32.650	1.933	.171[c]
	Residual	810.783	48	16.891		
	Total	843.433	49			
2	Regression	57.766	4	14.441	.827	.515[d]
	Residual	785.667	45	17.459		
	Total	843.433	49			

a. Dependent Variable: KValgusIn_45

b. Selecting only cases for which Exercise = 0

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	-2.315	2.496		-.928	.358			
	Tanner	.990	.712	.197	1.390	.171	.197	.197	.197
2	(Constant)	-1.641	7.625		-.215	.831			
	Tanner	1.224	.785	.243	1.559	.126	.197	.226	.224
	BJ	.018	.036	.075	.495	.623	.119	.074	.071
	VO2_est	-.042	.128	-.058	-.325	.747	.045	-.048	-.047
	BF	-.123	.127	-.179	-.967	.339	-.080	-.143	-.139

a. Dependent Variable: KValgusIn_45

b. Selecting only cases for which Exercise = 0

Regression: Post Exercise Initial Knee Valgus

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 1 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.287[a]	.082	.063	3.8956973	.082	4.304	1	48	.043
2	.337[b]	.113	.035	3.9547760	.031	.526	3	45	.667

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	65.326	1	65.326	4.304	.043[c]
	Residual	728.470	48	15.176		
	Total	793.795	49			
2	Regression	89.984	4	22.496	1.438	.237[d]
	Residual	703.811	45	15.640		
	Total	793.795	49			

a. Dependent Variable: KValgusIn_45

b. Selecting only cases for which Exercise = 1

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	-3.502	2.366		-1.481	.145			
	Tanner	1.400	.675	.287	2.075	.043	.287	.287	.287
2	(Constant)	-3.263	7.217		-.452	.653			
	Tanner	1.581	.743	.324	2.128	.039	.287	.302	.299
	BJ	.025	.034	.108	.729	.470	.150	.108	.102
	VO2_est	-.062	.121	-.090	-.515	.609	.011	-.077	-.072
	BF	-.112	.120	-.169	-.934	.355	-.031	-.138	-.131

a. Dependent Variable: KValgusIn_45

b. Selecting only cases for which Exercise = 1

Regression: Pre Exercise Peak Knee Valgus

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 0 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.055[a]	.003	-.018	7.1201060	.003	.148	1	48	.702
2	.347[b]	.120	.042	6.9076343	.117	1.999	3	45	.128

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	7.499	1	7.499	.148	.702[c]
	Residual	2433.404	48	50.696		
	Total	2440.902	49			
2	Regression	293.709	4	73.427	1.539	.207[d]
	Residual	2147.193	45	47.715		
	Total	2440.902	49			

a. Dependent Variable: PkValgus_45

b. Selecting only cases for which Exercise = 0

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	-7.561	4.323		-1.749	.087			
	Tanner	-.474	1.233	-.055	-.385	.702	-.055	-.055	-.055
2	(Constant)	-7.112	12.606		-.564	.575			
	Tanner	.133	1.298	.015	.102	.919	-.055	.015	.014
	BJ	.086	.060	.213	1.441	.157	.224	.210	.201
	VO2_est	-.209	.212	-.171	-.986	.330	.068	-.145	-.138
	BF	-.377	.209	-.325	-1.802	.078	-.262	-.259	-.252

a. Dependent Variable: PkValgus_45

b. Selecting only cases for which Exercise = 0

Regression: Post Exercise Peak Knee Valgus

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 1 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.001[a]	.000	-.021	7.2987483	.000	.000	1	48	.995
2	.316[b]	.100	.020	7.1526465	.100	1.660	3	45	.189

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	.002	1	.002	.000	.995[c]
	Residual	2557.043	48	53.272		
	Total	2557.045	49			
2	Regression	254.829	4	63.707	1.245	.306[d]
	Residual	2302.216	45	51.160		
	Total	2557.045	49			

a. Dependent Variable: PkValgus_45

b. Selecting only cases for which Exercise = 1

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	-9.120	4.432		-2.058	.045			
	Tanner	-.009	1.264	-.001	-.007	.995	-.001	-.001	-.001
2	(Constant)	-13.485	13.053		-1.033	.307			
	Tanner	.544	1.344	.062	.405	.687	-.001	.060	.057
	BJ	.075	.062	.181	1.212	.232	.223	.178	.171
	VO2_est	-.053	.219	-.043	-.243	.809	.150	-.036	-.034
	BF	-.313	.217	-.263	-1.441	.157	-.250	-.210	-.204

a. Dependent Variable: PkValgus_45

b. Selecting only cases for which Exercise = 1

Regression: Pre Exercise Valgus Excursion

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 0 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.219[a]	.048	.028	5.4474536	.048	2.408	1	48	.127
2	.382[b]	.146	.070	5.3283961	.098	1.723	3	45	.176

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	71.443	1	71.443	2.408	.127[c]
	Residual	1424.388	48	29.675		
	Total	1495.831	49			
2	Regression	218.200	4	54.550	1.921	.123[d]
	Residual	1277.631	45	28.392		
	Total	1495.831	49			

a. Dependent Variable: ValgusEx_45

b. Selecting only cases for which Exercise = 0

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	-5.245	3.308		-1.586	.119			
	Tanner	-1.464	.943	-.219	-1.552	.127	-.219	-.219	-.219
2	(Constant)	-5.471	9.724		-.563	.576			
	Tanner	-1.092	1.001	-.163	-1.090	.281	-.219	-.160	-.150
	BJ	.068	.046	.215	1.480	.146	.197	.215	.204
	VO2_est	-.167	.163	-.175	-1.023	.312	.053	-.151	-.141
	BF	-.255	.162	-.280	-1.578	.122	-.275	-.229	-.217

a. Dependent Variable: ValgusEx_45

b. Selecting only cases for which Exercise = 0

Regression: Post Exercise Valgus Excursion

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 1 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.203[a]	.041	.021	5.6710490	.041	2.056	1	48	.158
2	.348[b]	.121	.043	5.6075256	.080	1.365	3	45	.266

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	66.125	1	66.125	2.056	.158[c]
	Residual	1543.718	48	32.161		
	Total	1609.843	49			
2	Regression	194.847	4	48.712	1.549	.204[d]
	Residual	1414.995	45	31.444		
	Total	1609.843	49			

a. Dependent Variable: ValgusEx_45

b. Selecting only cases for which Exercise = 1

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	-5.618	3.444		-1.631	.109			
	Tanner	-1.408	.982	-.203	-1.434	.158	-.203	-.203	-.203
2	(Constant)	-10.222	10.233		-.999	.323			
	Tanner	-1.036	1.054	-.149	-.984	.330	-.203	-.145	-.137
	BJ	.050	.049	.152	1.032	.308	.175	.152	.144
	VO2_est	.009	.172	.009	.053	.958	.181	.008	.007
	BF	-.201	.170	-.212	-1.179	.245	-.294	-.173	-.165

a. Dependent Variable: ValgusEx_45

b. Selecting only cases for which Exercise = 1

Regression: Pre Exercise Initial Internal Tibial Rotation

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 0 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.012[a]	.000	-.021	8.7447640	.000	.006	1	48	.936
2	.228[b]	.052	-.032	8.7949630	.052	.818	3	45	.491

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	.494	1	.494	.006	.936[c]
	Residual	3670.603	48	76.471		
	Total	3671.097	49			
2	Regression	190.285	4	47.571	.615	.654[d]
	Residual	3480.812	45	77.351		
	Total	3671.097	49			

a. Dependent Variable: KRotIn_45

b. Selecting only cases for which Exercise = 0

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	-4.213	5.310		-.793	.431			
	Tanner	.122	1.514	.012	.080	.936	.012	.012	.012
2	(Constant)	-15.131	16.050		-.943	.351			
	Tanner	.250	1.652	.024	.151	.880	.012	.023	.022
	BJ	.094	.076	.189	1.232	.224	.208	.181	.179
	VO2_est	-.011	.270	-.008	-.042	.966	.101	-.006	-.006
	BF	-.148	.267	-.104	-.557	.581	-.125	-.083	-.081

a. Dependent Variable: KRotIn_45

b. Selecting only cases for which Exercise = 0

Regression: Post Exercise Initial Internal Tibial Rotation

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 1 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.067[a]	.005	-.016	9.0401427	.005	.220	1	48	.641
2	.327[b]	.107	.027	8.8438322	.102	1.718	3	45	.177

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	17.953	1	17.953	.220	.641[c]
	Residual	3922.761	48	81.724		
	Total	3940.713	49			
2	Regression	421.112	4	105.278	1.346	.268[d]
	Residual	3519.602	45	78.213		
	Total	3940.713	49			

a. Dependent Variable: KRotIn_45

b. Selecting only cases for which Exercise = 1

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	-7.673	5.489		-1.398	.169			
	Tanner	.734	1.566	.067	.469	.641	.067	.067	.067
2	(Constant)	-18.264	16.139		-1.132	.264			
	Tanner	1.114	1.662	.102	.670	.506	.067	.099	.094
	BJ	.129	.077	.250	1.679	.100	.280	.243	.236
	VO2_est	-.098	.271	-.063	-.362	.719	.114	-.054	-.051
	BF	-.304	.268	-.206	-1.133	.263	-.181	-.167	-.160

a. Dependent Variable: KRotIn_45

b. Selecting only cases for which Exercise = 1

Regression: Pre Exercise Peak Internal Tibial Rotation

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 0 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.129[a]	.017	-.004	8.0913192	.017	.818	1	48	.370
2	.452[b]	.204	.133	7.5184326	.187	3.531	3	45	.022

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	53.580	1	53.580	.818	.370[c]
	Residual	3142.533	48	65.469		
	Total	3196.113	49			
2	Regression	652.406	4	163.102	2.885	.033[d]
	Residual	2543.707	45	56.527		
	Total	3196.113	49			

a. Dependent Variable: KRotInPk_45

b. Selecting only cases for which Exercise = 0

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	3.941	4.913		.802	.426			
	Tanner	1.268	1.401	.129	.905	.370	.129	.129	.129
2	(Constant)	-18.292	13.720		-1.333	.189			
	Tanner	1.272	1.413	.130	.900	.373	.129	.133	.120
	BJ	.183	.065	.395	2.811	.007	.428	.386	.374
	VO2_est	-.036	.231	-.026	-.156	.877	.155	-.023	-.021
	BF	-.191	.228	-.144	-.838	.406	-.156	-.124	-.112

a. Dependent Variable: KRotInPk_45

b. Selecting only cases for which Exercise = 0

Regression: Post Exercise Peak Internal Tibial Rotation

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 1 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.092[a]	.008	-.012	8.9754817	.008	.410	1	48	.525
2	.398[b]	.159	.084	8.5387137	.150	2.679	3	45	.058

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	32.993	1	32.993	.410	.525[c]
	Residual	3866.845	48	80.559		
	Total	3899.838	49			
2	Regression	618.905	4	154.726	2.122	.094[d]
	Residual	3280.933	45	72.910		
	Total	3899.838	49			

a. Dependent Variable: KRotInPk_45

b. Selecting only cases for which Exercise = 1

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	5.175	5.450		.950	.347			
	Tanner	.995	1.554	.092	.640	.525	.092	.092	.092
2	(Constant)	-7.641	15.582		-.490	.626			
	Tanner	1.326	1.604	.123	.827	.413	.092	.122	.113
	BJ	.169	.074	.329	2.282	.027	.351	.322	.312
	VO2_est	-.184	.262	-.119	-.701	.487	.094	-.104	-.096
	BF	-.342	.259	-.233	-1.319	.194	-.184	-.193	-.180

a. Dependent Variable: KRotInPk_45

b. Selecting only cases for which Exercise = 1

Regression: Pre Exercise Internal Tibial Rotation Excursion

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 0 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.179[a]	.032	.012	6.6774674	.032	1.597	1	48	.212
2	.320[b]	.103	.023	6.6404462	.071	1.179	3	45	.328

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	71.192	1	71.192	1.597	.212[c]
	Residual	2140.251	48	44.589		
	Total	2211.443	49			
2	Regression	227.144	4	56.786	1.288	.289[d]
	Residual	1984.299	45	44.096		
	Total	2211.443	49			

a. Dependent Variable: KIRotEx_45

b. Selecting only cases for which Exercise = 0

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	7.281	4.055		1.796	.079			
	Tanner	1.461	1.156	.179	1.264	.212	.179	.179	.179
2	(Constant)	-5.609	12.118		-.463	.646			
	Tanner	1.264	1.248	.155	1.013	.317	.179	.149	.143
	BJ	.105	.058	.271	1.815	.076	.282	.261	.256
	VO2_est	-.046	.204	-.040	-.227	.822	.041	-.034	-.032
	BF	-.032	.201	-.029	-.160	.873	-.004	-.024	-.023

a. Dependent Variable: KIRotEx_45

b. Selecting only cases for which Exercise = 0

Regression: Post Exercise Internal Tibial Rotation Excursion

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 1 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.065[a]	.004	-.016	7.1820490	.004	.206	1	48	.652
2	.148[b]	.022	-.065	7.3512303	.018	.272	3	45	.845

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	10.623	1	10.623	.206	.652[c]
	Residual	2475.928	48	51.582		
	Total	2486.550	49			
2	Regression	54.724	4	13.681	.253	.906[d]
	Residual	2431.826	45	54.041		
	Total	2486.550	49			

a. Dependent Variable: KIRotEx_45

b. Selecting only cases for which Exercise = 1

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	11.970	4.361		2.745	.008			
	Tanner	.564	1.244	.065	.454	.652	.065	.065	.065
2	(Constant)	8.033	13.415		.599	.552			
	Tanner	.433	1.381	.050	.313	.756	.065	.047	.046
	BJ	.055	.064	.133	.857	.396	.119	.127	.126
	VO2_est	-.102	.225	-.083	-.452	.653	-.039	-.067	-.067
	BF	-.021	.223	-.018	-.094	.925	.021	-.014	-.014

a. Dependent Variable: KIRotEx_45

b. Selecting only cases for which Exercise = 1

Regression: Pre Exercise Relative Knee Energy Absorption

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 0 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.051[a]	.003	-.018	.14947	.003	.123	1	48	.728
2	.251[b]	.063	-.020	.14963	.060	.966	3	45	.417

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	.003	1	.003	.123	.728[c]
	Residual	1.072	48	.022		
	Total	1.075	49			
2	Regression	.068	4	.017	.755	.560[d]
	Residual	1.007	45	.022		
	Total	1.075	49			

a. Dependent Variable: RelKWA_45

b. Selecting only cases for which Exercise = 0

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	.371	.091		4.086	.000			
	Tanner	-.009	.026	-.051	-.350	.728	-.051	-.051	-.051
2	(Constant)	.112	.273		.411	.683			
	Tanner	-.008	.028	-.045	-.286	.776	-.051	-.043	-.041
	BJ	.002	.001	.192	1.260	.214	.215	.185	.182
	VO2_est	.001	.005	.055	.308	.759	.152	.046	.045
	BF	-.002	.005	-.072	-.388	.700	-.152	-.058	-.056

a. Dependent Variable: RelKWA_45

b. Selecting only cases for which Exercise = 0

Regression: Post Exercise Relative Knee Energy Absorption

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics				
	Exercise = 1 (Selected)				R Square Change	F Change	df1	df2	Sig. F Change
1	.133[a]	.018	-.003	.14682	.018	.871	1	48	.355
2	.143[b]	.020	-.067	.15144	.003	.039	3	45	.989

a. Predictors: (Constant), Tanner

b. Predictors: (Constant), Tanner, VO2_est, BJ, BF

ANOVA[a,b]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	.019	1	.019	.871	.355[c]
	Residual	1.035	48	.022		
	Total	1.053	49			
2	Regression	.021	4	.005	.234	.918[d]
	Residual	1.032	45	.023		
	Total	1.053	49			

a. Dependent Variable: RelKWA_45

b. Selecting only cases for which Exercise = 1

c. Predictors: (Constant), Tanner

d. Predictors: (Constant), Tanner, VO2_est, BJ, BF

Coefficients[a,b]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	Correlations		
		B	Std. Error	Beta			Zero-order	Partial	Part
1	(Constant)	.242	.089		2.710	.009			
	Tanner	.024	.025	.133	.933	.355	.133	.133	.133
2	(Constant)	.272	.276		.985	.330			
	Tanner	.027	.028	.152	.950	.347	.133	.140	.140
	BJ	2.109E-005	.001	.002	.016	.987	.027	.002	.002
	VO2_est	.000	.005	-.014	-.077	.939	.009	-.012	-.011
	BF	-.001	.005	-.060	-.316	.753	-.002	-.047	-.047

a. Dependent Variable: RelKWA_45

b. Selecting only cases for which Exercise = 1

Mixed Model Analysis: Initial Knee Valgus

Estimates of Fixed Effects[a]

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	-1.641001	7.423849	53.829	-.221	.826	-16.526003	13.244000
Tanner	1.224016	.764318	53.829	1.601	.115	-.308461	2.756493
BF	-.122567	.123377	53.829	-.993	.325	-.369942	.124807
BJ	.017936	.035306	53.829	.508	.614	-.052853	.088725
VO2_est	-.041661	.124746	53.829	-.334	.740	-.291780	.208459
Exercise	-1.622119	4.457692	45.000	-.364	.718	-10.600373	7.356134
Tanner_Exercise	.356898	.458939	45.000	.778	.441	-.567454	1.281249
BF_Exercise	.010566	.074083	45.000	.143	.887	-.138644	.159776
BJ_Exercise	.007075	.021200	45.000	.334	.740	-.035623	.049773
VO2_Exercise	-.020803	.074905	45.000	-.278	.782	-.171669	.130063

a. Dependent Variable: KValgusIn_45.

Mixed Model Analysis: Peak Knee Valgus

Estimates of Fixed Effects[a]

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	-7.112288	12.831078	58.466	-.554	.581	-32.792122	18.567545
Tanner	.132502	1.321017	58.466	.100	.920	-2.511352	2.776355
BF	-.377493	.213240	58.466	-1.770	.082	-.804267	.049282
BJ	.086366	.061021	58.466	1.415	.162	-.035760	.208493
VO2_est	-.208793	.215606	58.466	-.968	.337	-.640303	.222717
Exercise	-6.372700	9.351677	45.000	-.681	.499	-25.207944	12.462543
Tanner_Exercise	.411913	.962797	45.000	.428	.671	-1.527259	2.351086
BF_Exercise	.064948	.155416	45.000	.418	.678	-.248076	.377972
BJ_Exercise	-.011131	.044474	45	-.250	.804	-.100707	.078444
VO2_Exercise	.155438	.157141	45.000	.989	.328	-.161059	.471935

a. Dependent Variable: PkValgus_45.

Mixed Model Analysis: Knee Valgus Excursion

Estimates of Fixed Effects[a]

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	-5.471287	9.981597	60.560	-.548	.586	-25.433656	14.491082
Tanner	-1.091514	1.027650	60.560	-1.062	.292	-3.146729	.963701
BF	-.254925	.165885	60.560	-1.537	.130	-.586681	.076830
BJ	.068431	.047470	60.560	1.442	.155	-.026505	.163366
VO2_est	-.167133	.167725	60.560	-.996	.323	-.502569	.168304
Exercise	-4.750581	7.767398	45.000	-.612	.544	-20.394923	10.893761
Tanner_Exercise	.055016	.799688	45.000	.069	.945	-1.555639	1.665670
BF_Exercise	.054382	.129087	45.000	.421	.676	-.205612	.314376
BJ_Exercise	-.018207	.036940	45.000	-.493	.624	-.092607	.056194
VO2_Exercise	.176241	.130519	45.000	1.350	.184	-.086638	.439120

a. Dependent Variable: ValgusEx_45.

Mixed Model Analysis: Initial Internal Tibial Rotation

Estimates of Fixed Effects[a]

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	-15.130731	16.094364	49.347	-.940	.352	-47.467807	17.206345
Tanner	.250147	1.656987	49.347	.151	.881	-3.079100	3.579393
BF	-.148480	.267473	49.347	-.555	.581	-.685891	.388931
BJ	.094070	.076541	49.347	1.229	.225	-.059717	.247856
VO2_est	-.011423	.270441	49.347	-.042	.966	-.554798	.531952
Exercise	-3.133517	6.917383	45.000	-.453	.653	-17.065840	10.798807
Tanner_Exercise	.863363	.712175	45	1.212	.232	-.571032	2.297758
BF_Exercise	-.155402	.114960	45	-1.352	.183	-.386944	.076139
BJ_Exercise	.034767	.032897	45	1.057	.296	-.031492	.101025
VO2_Exercise	-.086773	.116236	45	-.747	.459	-.320884	.147338

a. Dependent Variable: KRotIn_45.

Mixed Model Analysis: Peak Internal Tibial Rotation

Estimates of Fixed Effects[a]

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	-18.292314	14.680693	51.107	-1.246	.218	-47.763536	11.178909
Tanner	1.271681	1.511443	51.107	.841	.404	-1.762513	4.305875
BF	-.191188	.243979	51.107	-.784	.437	-.680971	.298596
BJ	.183428	.069818	51.107	2.627	.011	.043270	.323585
VO2_est	-.035990	.246686	51.107	-.146	.885	-.531208	.459228
Exercise	10.651490	7.417474	45.000	1.436	.158	-4.288070	25.591050
Tanner_Exercise	.054611	.763662	45.000	.072	.943	-1.483484	1.592705
BF_Exercise	-.150424	.123271	45.000	-1.220	.229	-.398705	.097858
BJ_Exercise	-.014289	.035276	45.000	-.405	.687	-.085337	.056760
VO2_Exercise	-.147557	.124639	45.000	-1.184	.243	-.398593	.103480

a. Dependent Variable: KRotInPk_45.

Mixed Model Analysis: Internal Tibial Rotation Excursion

Estimates of Fixed Effects[a]

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	-5.609223	12.782996	56.643	-.439	.662	-31.210224	19.991778
Tanner	1.263616	1.316067	56.643	.960	.341	-1.372121	3.899353
BF	-.032251	.212441	56.643	-.152	.880	-.457715	.393214
BJ	.104593	.060793	56.643	1.720	.091	-.017159	.226344
VO2_est	-.046174	.214799	56.643	-.215	.831	-.476359	.384011
Exercise	13.641978	8.718917	45	1.565	.125	-3.918822	31.202778
Tanner_Exercise	-.831050	.897651	45	-.926	.359	-2.639013	.976913
BF_Exercise	.011191	.144900	45	.077	.939	-.280653	.303034
BJ_Exercise	-.049930	.041465	45	-1.204	.235	-.133445	.033584
VO2_Exercise	-.055824	.146508	45	-.381	.705	-.350906	.239258

a. Dependent Variable: KIRotEx_45.

Mixed Model Analysis: Relative Energy Absorption at the Knee

Estimates of Fixed Effects[a]

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	.112344	.274706	51.522	.409	.684	-.439016	.663704
Tanner	-.008034	.028282	51.522	-.284	.778	-.064798	.048731
BF	-.001760	.004565	51.522	-.386	.701	-.010923	.007403
BJ	.001636	.001306	51.522	1.252	.216	-.000986	.004258
VO2_est	.001415	.004616	51.522	.307	.760	-.007850	.010680
Exercise	.159921	.143166	45	1.117	.270	-.128430	.448272
Tanner_Exercise	.035068	.014740	45	2.379	.022	.005381	.064755
BF_Exercise	.000308	.002379	45	.130	.897	-.004484	.005100
BJ_Exercise	-.001615	.000681	45	-2.372	.022	-.002986	-.000244
VO2_Exercise	-.001774	.002406	45	-.737	.465	-.006619	.003072

a. Dependent Variable: RelKWA_45.