A Dissertation

entitled

The Effects of an OpenNI / Kinect-Based Biofeedback Intervention on Kinematics at the Knee During Drop Vertical Jump Landings: Implications for Reducing Neuromuscular Predisposition to Non-Contact ACL Injury Risk in the Young Female Athlete

by

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Submitted to the Graduate Faculty as partial fulfillment of the requirements for the Doctor of Philosophy Degree in Exercise Science

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The University of Toledo
December 2013
An Abstract of

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Introduction: The purpose of this study was to design and evaluate the validity and effectiveness of a prototype real-time Kinect-based biofeedback and screening system (KBBFSS) during drop vertical jump (DVJ) ACL injury prevention training in young female athletes. We hypothesized that KBBFSS would be both valid and reliable as compared with traditional MOCAP, and that a four-week intervention using KBBFSS would be effective at improving landing kinematics. Methodology: 24 female gymnasts were randomized into control (CTRL) or Kinect-based biofeedback (KBF) groups. Eight of the subjects were additionally randomized into a validation subset. Subjects were grouped as “high risk” or “normal risk” using a novel risk stratification algorithm. Custom KBBFSS software afforded on-screen representation of limb and joint segments responding intuitively and immediately to subject movement. Subjects performed twenty 30cm drop landings three days per week for four weeks, wherein KBF subjects used the KBBFSS to augment landing mechanics, while CTRL subjects did so without KBBFSS. Alpha-level was set a priori at $p \leq 0.05$. 

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Results: KBBFSS results were valid for pre (r=0.963) and post (r=0.897) knee flexion, and pre (r=0.815) and post (r=0.916) knee separation distance as compared with MOCAP. Knee flexion change score was statistically different between groups (p=0.001) and effect size was large (d=1.618), power of 0.93. Knee separation distance change score was statistically different (p=0.024) between groups, with moderate effect size (d=0.99) and power of 0.73. KBF group reduced peak vGRF more than controls, with large effect size (d=1.84). KBF decreased peak bilateral frontal plane valgus knee moment more than controls, with moderate effect size (d=0.44). Correlations between pre-training RQS and changes in knee flexion and separation distance for “high risk” subjects were moderate. Conclusion: KBBFSS kinematic values are valid and KBF intervention significantly improved non-contact ACL injury risk knee kinematics. The RQS algorithm moderately predicted outcome measures, supporting previously established postulations that individuals who are at greatest functional risk of non-contact ACL injury stand to gain the greatest benefit from intervention. Though further research is warranted, in particular longitudinally, this new clinically-deployable tool may be effective in combating non-contact ACL injury in female adolescent athletes.
For Jack and Claire. May your thirst for knowledge never be satisfied and your academic pursuits know no limits.
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List of Abbreviations

3D .................. Three dimensional coordinate system
ACL .................. Anterior Cruciate Ligament of the knee
BW .................. Relative/normalized multiples of body weight
CTRL .................. Control group
Fx, Fy ................. Anterior-posterior and medial-lateral force vector components
GRF .................. Ground reaction force (Fz)
GUI .................. Graphical user interface
KBBFSS ............... Kinect-Based Biofeedback and Screening System
KBF .................. Kinect-based biofeedback training group
Knee Valgus ............ Refers to shank segment abducted laterally away from the midline of the body with reference to the thigh segment.
MOCAP ............... 3D Motion Capture
NI .................. Natural Interface
OpenNI ............... Open Natural Interface software platform
Chapter 1

Introduction

1.1 Epidemiology of ACL Injury

It is estimated that between 50,000 and 100,000 ACL injuries occur every year (Grindstaff, Hammill et al. 2006, Kramer, Denegar et al. 2007, Sugimoto, Myer et al. 2012a) with nearly 38,000 cases occurring in women (Hughes and Watkins 2006, Chaudhari, Lindenfeld et al. 2007). Nearly 50,000 reconstruction surgeries are performed in the US alone each year (Sugimoto, Myer et al. 2012a) with an average medical cost of $17,000 to $25,000 per case (Grindstaff, Hammill et al. 2006, Hewett, Myer et al. 2006a, Sugimoto, Myer et al. 2012a), yielding an estimated aggregate total of between 1 and 3 billion dollars per year (Hughes and Watkins 2006, Kramer, Denegar et al. 2007, Donnelly, Lloyd et al. 2012, Sugimoto, Myer et al. 2012a).

Of the aforementioned totals, nearly 70% of cases are reported to be non-contact etiology in nature (McNair, Marshall et al. 1990, Agel, Arendt et al. 2005). It appears that non-contact ACL injury rates have continued to rise, especially among female athletes (Borotikar, Newcomer et al. 2008).

As more young women become active in competitive sports, it is anticipated that the number of ACL injuries in women is going to continue to grow (Ford, Myer et al. 2003). While an injury to the ACL is a serious problem that generally disrupts an athlete’s participation in her chosen sport, such injury may have significant long term consequences as well. Many individuals who have experienced ACL injury, whether surgically corrected or not, experience some degree of knee osteoarthritis in later life, in some cases necessitating knee replacement surgery. In fact, it is estimated that more than half of ACL-injured knees will develop osteoarthritis as little as 10 years following the initial injury (Hootman and Albohm 2012). Thus, ACL injury is a serious and unrelenting issue for those involved in sports, and has profound implications financially and in terms of quality of life, for many individuals.
1.2 Structure and Function of the ACL

The anterior cruciate ligament (ACL) of the knee is primarily responsible for providing stability to the tibiofemoral joint (the knee), including: (1) preventing excessive sagittal plane anterior translation of the proximal tibia on the distal femur; (2) and aiding in the prevention of excessive transverse plane external rotation of the tibia relative to the femur (Griffin, Albohm et al. 2006, Hughes and Watkins 2006).

The engineered structure of the ACL consists of two distinct ligament bundles (Hara, Mochizuki et al. 2009). These bundles are the referred to as the anterior-medial (AM) and posterior-lateral (PL) bundles. Each has a slightly different orientation and thus “checks” against slightly different force vectors associated with their anatomical alignment (Hughes and Watkins 2006). Regardless of orientation, each bundle attaches proximally at the distal femur and distally at the proximal tibia (Hara, Mochizuki et al. 2009). It is also important to note, from a functional biomechanics perspective, that the relative orientation of the bundles is parallel in knee extension but slightly twisted upon one another in knee flexion (Hughes and Watkins 2006).

The histological structure of the ACL consists predominantly of collagen fibers, of a wavy shape, running both parallel and axially. The orientation of the ultrastructure provides for strength in multiple axes. Vascularization of the ACL is provided predominantly by the middle genicular artery, though some branches of the inferior geniculate artery may perfuse the distal component of the ligament (Toy, Yeasting et al. 1995). Notable is the greater vascularization of the outer components of the ligament, while the middle portion is significantly less perfused. Likewise, the proximal portions
of the ACL receive more blood flow than the distal portions. Neural innervation to the ACL is via posterior knee joint capsule penetration of the tibial nerve (Hara, Mochizuki et al. 2009). Additionally, local presence of Ruffini and Pacini corpuscles, and adjacent muscle spindles and golgi tendon organs (GTOs) provide afferent position information back to the central nervous system (CNS) with regards to sensorimotor control of knee positioning (Solomonow and Krogsgaard 2001, Hara, Mochizuki et al. 2009).

With regard to physical size and strain properties of the ACL, it is notable that the female ACL is significantly smaller, on average, than in males (White, Lee et al. 2003, Hughes and Watkins 2006) and is structurally capable of withstanding less strain (Quatman and Hewett 2009).

1.3 Mechanisms of Non-Contact ACL Injury

Efforts to reduce the rate of ACL injury require a complete understanding of the mechanism of injury. While it has been well established that the ACL can be ruptured under both contact and non-contact conditions, the scope of this paper will focus entirely on non-contact mechanisms of injury. Most non-contact ACL injuries occur during sudden deceleration, rapid change of direction, or while landing from a jump (Paszkewicz, Webb et al. 2012). It is under these dynamic conditions that the ACL appears to be most vulnerable from a biomechanical perspective.

Biomechanically, the ACL, which is responsible, as outlined above, for prevention of excessive anterior translation of the tibia on the femur in the sagittal plane
and for prevention of excessive transverse plane rotation at the knee joint, is vulnerable in positions of low knee flexion (near extension) and dynamic valgus (frontal plane) torque (Hewett, Myer et al. 2006b). Additionally, excessive transverse plane rotation, in conjunction with either decreased knee flexion and/or dynamic valgus, can be highly detrimental to the integrity of the ACL (Hewett, Myer et al. 2005).

It is important to note that the mechanism of non-contact ACL injury appears to be significantly different for males as compared with females. Males have a greater disposition to utilizing their knees like a hinge-joint, whereas females are much more prone to lower extremity movement patterns wherein the knees are utilized more like a loose ball and socket joint (Hewett, Myer et al. 2004, Ford, Myer et al. 2005a, Hewett, Zazulak et al. 2005). Additionally, females have also been shown to be more quadriceps dominant with regards to muscular strength (Hewett, Myer et al. 2008). The effect of the quadriceps dominance is decreased knee flexion, yielding decreased force amortization capability as well as known structural vulnerability of the ACL as a function of decreased knee flexion (Quatman and Hewett 2009). Likewise, decreased hamstrings activation upon, or immediately before, completing a jump landing has been implicated as a contributing factor in neuromuscular-linked failure of the ACL (Hewett, Myer et al. 2008, Medina, Valovich McLeod et al. 2008). Particularly in female athletes, dynamic knee valgus has been implicated as a major contributing factor to non-contact ACL injury (Hewett, Myer et al. 2006b, Quatman and Hewett 2009).

ACL injury risk factors are often characterized into two separate groups: extrinsic factors and intrinsic factors. Extrinsic factors are those that are external to the structural anatomy and are potentially modifiable. Such extrinsic factors include equipment
(athletic shoes, cleats, and field or court coefficient of friction) and training factors (muscular strength and neuromuscular coordination). Intrinsic factors are those that are not, or at least not easily, modifiable, such as anatomical intracondylar notch width, ACL size, limb alignment, and joint laxity (Arendt and Dick 1995).

Although some controversy exists over the exact mechanisms of dynamic non-contact ACL injury in females, it has been strongly contended that the functional risk is multifactorial in nature, incorporating both extrinsic and intrinsic factors (Hewett, Myer et al. 2006a, Shultz 2008, Quatman and Hewett 2009). It is likely that no one mechanism is independently responsible for non-contact ACL injury, but rather a combination of factors. Such factors include, but are not limited to, excessive dynamic valgus, decreased knee flexion, increased transverse plane torsion, imbalance of hamstring to quadriceps strength ration, tibial slope, and decrease mitigation of ground reaction forces (Shultz 2008).

While ACL injury is a multi-factorial problem, a recent trend for both ACL researchers and clinicians who work to prevent ACL injury, is to focus primarily on the factors that are most modifiable (Myer, Ford et al. 2005b). Anatomical factors related to gender may in fact predispose the athlete to injury, but these are not generally modifiable. However, neuromuscular control, strength-related factors, and motor learning factors are all indeed modifiable, Additionally, recent research has shown that through modification of these factors, in at least some instances, non-contact ACL injury rates can be decreased (Sugimoto, Myer et al. 2012b, Hewett, Di Stasi et al. 2013).
Approaches to ACL injury prevention in female athletes that have received recent attention are interventions that focus on diminishing dynamic valgus. Training interventions that involve feedback modulation of landing and cutting, designed to optimize ACL-related biomechanics, have shown significant promise in reducing biomechanics consistent with increased ACL injury risk. In a study published in 2006, Myer, and colleagues successfully demonstrated the benefits of feedback modulated jump landing training on reducing risky frontal plane biomechanics (valgus) consistent with increased risk of non-contact ACL injury (Myer, Ford et al. 2006).

In order to effectively address neuromuscular issues pertinent to ACL injury, quantification of the involved movements is necessary. Both the kinematic and kinetic components related to neuromuscular control can, with today’s modern biomechanical instrumentation, be tracked and studied with high precision. With the knowledge now established linking neuromuscular control to potential screening methodologies for ACL injury risk, the advances in research into non-contact injury mechanisms and the value of intervention, utilization of accurate quantitative measurement systems has become paramount. However, there are some significant limitations that act as a counter-balance against the advantages of such systems, including the high cost of the equipment (as much as half a million dollars for a complete system), the steep learning-curve for a clinician to be trained properly in its use, and the limited portability of such a system. Logically, the need for a system that provides the advantages of contemporary biomechanics instrumentation, without the significant limitations aforementioned, would be ideal for field and clinical use, thus bringing the benefits to many more athletes at risk of injury.
1.4 Statement of the Problem

Aberrant neuromuscular control, generally defined as a diminished functional capacity to control perturbation of one’s center of mass during human movement, has been identified as a contributing factor in negatively altered lower extremity biomechanics (Hughes, 2006). Excessive knee dynamic valgus during drop vertical jump (DVJ) landings is such a neuromuscular factor for increased risk of non-contact ACL injury in the female athlete (Ford, Myer et al. 2003, Hewett, Myer et al. 2004). While evidence has shown that targeted neuromuscular training interventions have the potential to reduce injury risk, successful clinical interventions are limited (Hewett, Lindenfeld et al. 1999, Myer, Stroube et al. 2013). While it has been established that feedback is an integral component in improving neuromuscular control related to knee function (Brindle, Mizelle et al. 2009, Benjaminse, Lemmink et al. 2010), employment of a real-time biofeedback training system has not been successfully implemented at the clinical level, due to numerous limitations including lack of capital expenditure funding, physical space, and lab time scheduling constraints for employment of effective interventions within a clinical environment.
1.5 Statement of the Purpose

The purpose of this study was two-fold: 1.) to design and evaluate a prototype Kinect-based biofeedback and screening system (KBBFSS) as a tool for providing real-time objective feedback of a subject’s kinematics during drop vertical jump (DVJ) ACL injury prevention training; and 2.) to determine the validity of this system in screening for lower extremity biomechanics consistent with risk of non-contact ACL injury predisposition during drop vertical jump landings in young female athletes.

1.6 Research Hypotheses

We hypothesized that the Kinect-based biofeedback and screening system (KBBFSS) would produce valid kinematic values as compared with a “gold-standard” 3D motion-capture (MOCAP) lab system (H1). We also hypothesized that the KBF intervention drop vertical jump (DVJ) training group would adopt greater knee flexion and greater knee separation values, following four weeks of training, as compared with the traditional CTRL DVJ training (non-biofeedback) group (H2). Additionally, we hypothesized that the KBF training group would reduce peak vertical ground reaction forces and frontal plane dynamic valgus moment, following four weeks of training as compared with the CTRL DVJ training (non-biofeedback) group (H3). Finally, we hypothesized that those subjects who were pre-training risk stratified (RQS) as “high-risk” would experience greater increases in knee flexion angle and knee separation distance than those stratified as “normal-risk” (H4).
1.7 Variables

For the intervention effect component, the experimental (independent) variable was the intervention training program ("group"). This was a categorical variable with two levels: (1) a group (KBF) that completed a four week program of drop-landings with real-time visual and auditory biofeedback intervention training program, and (2) a group (CTRL) that completed four week control group that performed the same drop landings, but without the feedback system.

The outcome (dependent) variables for the clinical evaluation (KBBFSS derived kinematics) were: peak knee flexion angle (degrees), and normalized peak minimum knee separation distance (mm/m).

The outcome (dependent) variables for the validation component (MOCAP lab) were: peak knee flexion angle (degrees), peak knee valgus joint moment (newton-meters), peak vertical ground reaction force, and peak minimum knee separation distance. All of the kinematic and kinetic dependent variables were of continuous statistical nature.
Chapter 2

Literature Review

2.1 Introduction

From early through contemporary studies, ACL injury has been deeply and broadly studied. Herein we will examine the literature relative to the anatomical/structural, hormonal, and proprioceptive qualities as well as neuromuscular control and neuromuscular timing of the knee specifically with relation to the predisposition to, and mechanism of, injury to the ACL in the female athlete. We will also examine previous and contemporary literature with regards to injury risk screening and intervention training programs aimed at reducing non-contact ACL injury rates. Additionally, we will explore the contemporary literature relative to OpenNI and Kinect / PrimeSense software and hardware components.
2.2 Anatomical Predisposition

For many years, researchers have grappled with the question of whether or not there are clearly definable physiological characteristics that delineate the potentially “at risk” knee (Chappell, Yu et al. 2002, Uhorchak, Scoville et al. 2003, Kramer, Denegar et al. 2007, Shultz 2008). While there have been many studies and much debate, it must be noted that the research has shifted in paradigm over to areas that are more readily modifiable, as most of the purely physiological characteristics (i.e.: Q-angle, femoral notch width, generalized joint laxity) are either not modifiable or not easily or cost/time-effectively modifiable (Hughes and Watkins 2006). However, due to the significant number of studies that have been conducted to evaluate these characteristics, it is relevant in the scope of this comprehensive literature review to present the most pertinent of these studies.

In 2007, Kramer, et al. (Kramer, Denegar et al. 2007) conducted a case control study of 33 females with a history of ACL injury and compared injury history as well as a number of lower extremity physiological measures between this group and a group of 33 female controls (non-ACL injured). The results of their study showed a trend between “generalized joint laxity” and ACL injury but did not indicate that any one specific physiological angle or structure could be identified as being directly linked to ACL injury predisposition (Kramer, Denegar et al. 2007). Limitations to this study are notable, specifically the small cohort size and the fact that the study evaluated subjects after having suffered an ACL injury, rather than before.
In 2006, Hughes and Watkins (Hughes and Watkins 2006) published an analysis of risk-factor modeling for ACL injury. In their findings, they noted that current models of etiology of injury fail to explain the interaction between various risk factors that contribute to injury. In examining risk factors individually, the ‘bigger picture’ is lost. Hughes and Watkins created a flow diagram of what they consider to be the major factors with relation to the passive stability of the tibiofemoral joint. Additionally, Hughes and Watkins evaluated these passive stability factors in conjunction with dynamic stability factors and developed their proposed risk factor model. As a result of their investigation, Hughes and Watkins postulated that sex differences in dynamic (rather than passive physiological) stabilization are responsible for the increased risk of non-contact ACL injury in female athletes (Hughes and Watkins 2006).

In 2008, at the request of the International Olympic Committee (IOC), Renstrom and colleagues examined current injury mechanism studies and related evaluations of ACL injury in female athletes (Renstrom, Ljungqvist et al. 2008). In their published report, they looked closely at internal and external risk factors for non-contact ACL injury. With respect to the internal risk factors (the more pertinent of the two to this particular literature review), the researchers found that although there is some evidence to suggest an association between intracondylar notch width and injury, this trend is likely linked to the causal relationship between notch size and the size of the ACL. In this case, because women, in general, have small notches, they also have smaller ACLs. In summarizing their review of internal risk factors, they note that anatomical risk factors, although possible, are likely not easy to correct (Renstrom, Ljungqvist et al. 2008). As a
result, understanding predisposition from a wider perspective in order to screen out and intervene in specific cases becomes a more relevant approach.

In 2000, Hewett published a review that examined factors associated with knee injuries in female athletes. In his review, Hewett discussed three “theories” within the industry at that time, with regards to increased knee injury in female athletes. These theories are: anatomical, neuromuscular, and hormonal. Specifically with regard to the anatomical theory, Hewett examined factors thought to be related to ACL injury; in this case, the quadriceps angle (Q-angle). Hewett referenced studies that both did and did not find a direct relationship to support this theory. As a result, he concluded that the results are contradictory. In addition, any proposed link between this anatomical / physiological predisposition and ACL injury is not normally modifiable. As a result, Hewett recommended, at that time, that more focus and effort be placed on the hormonal and neuromuscular theories.

A review of potential anatomical predisposition to ACL injury in female athletes would not be complete without elaborating on the research that has been conducted on generalized joint laxity and its relationship to ACL injury predisposition. A number of studies have been conducted in this area (Rozzi, Lehpert et al. 1999, Ramesh, Von Arx et al. 2005, Myer, Ford et al. 2008, Quatman, Ford et al. 2008). Herein is a brief review of some the studies reviewed as part of this particular literature review:

In Ramesh, et al. (Ramesh, Von Arx et al. 2005), researchers evaluated the relationship between generalized joint laxity and ACL injury. Over a period of two years, they evaluated generalized joint laxity in each of 169 patients undergoing ACL
reconstruction surgery. The results of their investigation yielded a correlation between generalized joint laxity (particularly hyperextension of the knee, otherwise referred to as genu recurvatum) and ACL injury occurrence.

Quatman, et al. (Quatman, Ford et al. 2008) published a more modern study of generalized ligament laxity with regards to the effects of gender and pubertal status and the ultimate potentiality for increased injury risk. In their study, 275 female and 143 male subjects were assessed for generalized joint laxity (as prescribed utilizing the Beighton and Horan Joint Mobility Index). As a result of the study, the researchers determined that females displayed increased joint laxity scores between pre and post pubertal groups while males did not. Interestingly, pre-pubertal male and female scores were collectively similar. It was not until after the onset of puberty that the disparity became pronounced. In concluding their study, the research group surmised that the structural and physiological changes that occur through pubertal development (in this case, specifically passive joint restraints) may contribute to the incidence of injury.

The final study reviewed with respect to generalized joint laxity was conducted and published by Myer, et al. in 2008. In their study, 1558 female soccer and basketball players were prospectively screened with 19 going on to tear their ACL. In the particular data analyses conducted, the researchers compared these 19 ACL injured subjects’ data to 76 height and mass-matched control subjects, selected out of the original group, for generalized joint laxity and anterior-posterior tibiofemoral translation (utilizing a CompuKT knee arthrometer). The results of their study indicated that increased knee laxity measures may contribute to increased risk of ACL injury. Additionally notable,
they found that a positive measure of knee hyperextension (genu recurvatum in the sagittal plane) increased the odds of ACL injury status five-fold (Myer, Ford et al. 2008).

2.3 Hormonal Contributions


Beynnon, et al. published a study of the effect of estradiol and progesterone on knee and ankle joint laxity. In their study, 17 healthy female athletes and 17 male controls were evaluated for ankle laxity, anterior-posterior knee joint laxity, and both estradiol and progesterone blood serum levels. As a result of the study, it was determined that although the female subjects as a whole had greater ankle and knee joint laxities than the males, there was no evidence of a change in laxity over the course of the menstrual cycle (Beynnon, Bernstein et al. 2005).
Friden, et al. (Friden, Hirschberg et al. 2006) published a 2006 study which measured knee joint kinesthesia and neuromuscular coordination against different phases of the menstrual cycle of moderately active females. In their study, subjects (32 females), underwent the same neuromuscular coordination and kinesthesia testing process, specifically a 30-second square hop test, in each of the three delineated phases of the menstrual cycle. The results showed that the variation in hormone levels through the menstrual cycle yielded performance effects on knee joint neuromuscular coordination and kinesthesia.

In 2007, Hewett, et al. published a systematic review in which prominent previously published studies were examined in order to collectively determine trends in the findings, specifically with regards to ACL injury. Overall, the seven studies that they reviewed showed a collective favor for an evident effect of menstrual cycle hormonal fluctuation in the first half (pre-ovulatory phase) with respect to greater incidence of ACL injury (Hewett, Zazulak et al. 2007).

Again in 2007, Abt, et al. produced a study on the effects of menstrual cycle phase with respect to neuromuscular and biomechanical functional characteristics in physically active females. In this small cohort study (only ten females), two blood hormone levels (estradiol and progesterone) were obtained and nine other neuromuscular and biomechanical tests were conducted utilizing force plate and 3D motion analysis systems. In concluding their study, this research group found no clear relationship between female blood hormone levels and neuromuscular or biomechanical characteristics of movement (Abt, Sell et al. 2007).
Agel, et al. (Agel, Bershadsky et al. 2006) published a 2007 study examining hormone therapy effects on ACL and ankle injuries in female NCAA basketball and soccer athletes. In their study, the effects of oral contraception on incidence of injury were evaluated. From a sample size of 3150, approximately 1/3 were oral contraceptive users and 2/3 were non-users. In the study, the researchers tracked injury rates and plotted the injury incidence against use of hormone therapy (oral contraception). As a result of their study, the researchers found no difference in injury rate between oral contraception users and non-users.

Chaudhari, et al. (Chaudhari, Lindenfeld et al. 2007) evaluated knee and hip loading patterns during different phases of the menstrual cycle. Although of small cohort size (25 women), this study did have relatively equal numbers of oral-contraceptive users (13) and non-users (12), as well as 12 male controls. In this study, subjects were evaluated via 3D motion analysis kinematics while performing horizontal, vertical, and drop-landing tasks. Male controls were tested once and females were tested twice each for each of the three phases of the menstrual cycle (follicular, luteal, and ovulatory). As a result of the study, the researchers determined that menstrual cycle hormone level fluctuation and contraceptive use had no effect on knee or hip loading patterns.

The final study that was reviewed with respect to the relationship between female hormone levels and ACL injury was published in 2008 by Renstrom, et al. In their ‘wide-cast’ review study (encompassed all internal and external potential factors) supported by the International Olympic Committee (IOC), the researchers found that there was no conclusive evidence that support a benefit of female hormone regulation/therapy (oral contraception) on reducing ACL injury rate (Renstrom, Ljungqvist et al. 2008).
In summary, the effects of female hormone levels on ACL integrity and risk of injury are inconclusive. Although there have been a number of studies to suggest a higher rate of injury in the ‘first-half’ (pre-ovulatory phase) of the menstrual cycle, (Friden, Hirschberg et al. 2006, Renstrom, Ljungqvist et al. 2008) there are also some inconclusive studies that refute and confound a clear delineation of ‘time in the cycle’ compared with injury risk (Abt, Sell et al. 2007, Hewett, Zazulak et al. 2007, Adachi, Nawata et al. 2008). Additionally, although there have been many studies to examine the effects of hormone therapy (oral contraception) on risk of ACL injury (Agel, Bershadsky et al. 2006, Hewett, Zazulak et al. 2007, Renstrom, Ljungqvist et al. 2008), there are contradicting results which lead to no clear conclusion. What does appear clear is that, while there may be some contribution to injury risk provided by fluctuating female hormone levels, the more significant factors are related to neuromuscular, proprioceptive, and strength components (Chaudhari, Lindenfeld et al. 2007, Hewett, Zazulak et al. 2007, Renstrom, Ljungqvist et al. 2008).

2.4 Muscular Strength and Muscular Recruitment

In reviewing the kinetic and kinematic studies, it is impossible to complete a thorough literature review without examining the plethora of research relating to the issue of motor recruitment strategies and muscular strength imbalances. Many studies have examined the differences in motor unit recruitment and muscular strength imbalances between female and male athletes (Cowling and Steele 2001, White, Lee et al. 2003,
Cowling and Steele, in 2001, published a study that evaluated lower extremity muscle synchrony during landing as it relates to gender as well as the potential implications on ACL injury rates (Cowling and Steele 2001). Although the study cohort size was small (7 males, 11 females), this study utilized 2D kinematic analysis, force plate technology, and EMG data sampling to determine the effect of muscle synchrony and gender on risk of ACL injury. Participants caught a net-ball pass and subsequently landed in single leg stance (balance) on the force plate. Upon analyzing the data obtained, the researchers concluded that female subjects demonstrated less controlled (protected) knee joint muscle activity than males, therein potentially increasing their risk of non-contact ACL injury as a result.

In 2003, White, et al. conducted and published a study that examined differences between female and male athletes’ hamstring and quadriceps activity during dynamic, fatiguing exercise. Utilizing EMG power spectra, the researchers determined MVC for knee flexion and extension in consecutive two-minute bouts of exercise. As a result, the group concluded that female athletes exhibited significantly higher quadriceps co-activation patterns leading to a greater risk of potential non-contact ACL injury as a result of higher anterior tibial loads experienced at comparatively reduced angles of knee flexion (White, Lee et al. 2003).

Myer, et al. (Myer, Ford et al. 2005a) conducted a study, in 2005, that evaluated quadriceps muscle activation via a mimicked high risk of ACL injury maneuver in order
to determine gender disparity. In their study, subjects (ten males and ten females) performed lateral (frontal plane) weight shifting exercises at predetermined velocities. EMG data collection of the medial and lateral quadriceps was conducted and analysis was performed. As a result, the researchers found that the female subjects demonstrated decreased medial to lateral quadriceps ratios compared with males, and resultantly have a higher risk of ACL injury, due specifically to the increase in “dynamic valgus” experienced (Myer, Ford et al. 2005b).

In Hughes and Watkins’ 2006 publication, they examined risk-factors for ACL injury and noted that females tend to produce a quadriceps-dominant mode of stabilizing the knee joint which may increase the risk for ACL injury. Additionally, the researchers noted that the speed of muscle reaction time, as well as time to peak torque are factors in protecting the ACL, dynamically. In general, males demonstrated shorter reaction times producing less resultant strain on the ACL. Finally, they also reviewed muscle stiffness and strength as a factor, and concluded that muscle stiffness (resulting from muscular strength) in the hamstrings and quadriceps, both, may contribute to increased risk of ACL injury (Hughes and Watkins 2006).

More recently, in 2008, Hewett, et al. examined hamstring to quadriceps peak torque ratios with increasing isokinetic angular velocity between males and females. In their analysis of a large amount of data (1568 subjects), it was noted that males demonstrated a significant correlation between hamstring to quadriceps (H/Q) ratio and isokinetic velocity, as well as a significant difference in resulting H/Q ratio between low and high angular velocities, which was not apparent in female subjects. As a result, it was concluded that females, unlike their male counterparts, did not increase their H/Q
torque ratios at or near functional velocities, yielding again a greater potential for injury to the ACL (Hewett, Myer et al. 2008).

Finally, and most recently, in 2009, Myer, et al. published a study that evaluated the relationship of hamstrings and quadriceps strength to ACL injury in female athletes (Myer, Ford et al. 2009). In their study, subjects (22, with 88 controls) were evaluated in performing isokinetic knee extension and flexion at 300 degrees per second. As a result, the researchers concluded that female athletes who later succumbed to ACL injury yielded test results indicative of decreased hamstrings strength but equivalent quadriceps strength when compared with male controls. Interestingly, this was in contrast to the data obtained for female athletes who did not go on to ACL injury. These individuals had less quadriceps strength but equivalent hamstring strength when compared to the male participants.

In summary, it is evident that the ratio of quadriceps to hamstring strength is a clear component when considering the risk of non-contact ACL injury in female athletes (White, Lee et al. 2003, Hughes and Watkins 2006, Hewett, Myer et al. 2008, Myer, Ford et al. 2009). It appears reasonable that strength training and plyometric activity may help to correct, to at least some degree, this imbalance and thus provide a potentially reduced risk of non-contact ACL injury in the female athlete (White, Lee et al. 2003, Hughes and Watkins 2006, Myer, Ford et al. 2009).
2.5 Proprioceptive Modulation and Neuromuscular Influences


In 2004, Hewett, et al. (Hewett, Myer et al. 2004) evaluated biomechanical measures of neuromuscular control with additional respect to valgus loading of the knee as predictors for ACL injury risk in female athletes. In their study, 205 female athletes were kinematically and kinetically evaluated during a jump-landing maneuver. In this study, it was evident that athletes with a confirmed ACL tear exhibited significantly different knee joint biomechanics and loading as compared with those who did not experience an ACL injury. More precisely, the knee joint abduction angle, upon landing, was eight degrees greater in the ACL injured athletes. The ACL injured subjects also experienced 2.5 times greater knee abduction moment (torque) and 20% greater ground reaction force. In this study, knee joint abduction moment predicted ACL injury with
73% specificity, leading to the conclusion that knee motion together with knee loading during landing are predictors of ACL injury in female athletes.

As a component of their 2006 review of risk factors for ACL injury, Hughes and Watkins (Hughes and Watkins 2006) examined the effects of fatigue on dynamic stability and resultant risk of ACL injury. In their findings, they point out that muscle fatigue alters neuromuscular reaction to the anterior translation of the tibia on the femur, therein reducing dynamic stability.

Sell, et al. investigated the predictors of proximal tibia anterior sheer force during the vertical stop-jump. In their study, 36 high school athletes (19 males and 17 females) were evaluated performing vertical stop-jump control tasks utilizing EMG, force plate, and 3D motion analysis systems. The results of the kinematic study indicated that knee flexion moment (torque) had the greatest effect on proximal tibia anterior shear force (Sell, Ferris et al. 2007).

Zazulak, et al., in 2007, studied the effects of core proprioception on knee injury. In their study, researchers subjected 277 collegiate athletes (approximately half male and half female) to passive and actively-repositioned core position perturbation. As a result of the study, it was evident that impaired proprioception predicted knee injury risk in female, but not male, athletes (Zazulak, Hewett et al. 2007b).

In 2008, Borotikar, et al. (Borotikar, Newcomer et al. 2008) published their study of the combined effects of central and peripheral contributions to ACL injury risk. In this study, they combined imposition of fatigue and decision making variables during sports landing conditions. Twenty-five female NCAA athletes were evaluated, using 3D
kinematics, for initial and peak stance phase on landing under anticipated and unanticipated conditions, before and after accumulation of fatigue. Their findings indicated that combined fatigue and altered decision making processes present a “worst-case” scenario wherein overload and degradation of central control mechanisms lead to greater risk of ACL injury by increasing contact phase hip extension and internal rotation as well as peak stance knee abduction.

Zazulak, et al., again in 2007, conducted a study of 277 collegiate athletes (approximately half male, half female) to evaluate unanticipated trunk displacement and the effects of this perturbation on knee injury risk. In the study, the subjects’ trunk was displaced via sudden release (electromagnetic) of unilateral resistance. Via a “flock of birds” electromagnetic device, resultant displacement of the trunk was measured. As a result of the data recorded, the researchers concluded that factors related to core stability predicted ACL injury with high sensitivity and moderate specificity in female athletes (Zazulak, Hewett et al. 2007a).

Utilizing a relatively small cohort (18 high school female athletes), Myer, et al. (Myer, Ford et al. 2007) conducted a 2007 study examining differential neuromuscular training effects on ACL injury risk, specifically with regards to classification of “high-risk” and “low-risk” female athletes. Subjects were initially evaluated (via 3D motion analysis of their knee abduction moments upon landing in a drop vertical jump task) for placement into “high risk” (>25.25 Nm) and “low risk” (<25.25 Nm) groups. Following seven weeks of intervention based training, the groups were re-evaluated. Resultant data showed that the “high risk” athletes successfully reduced the magnitude of knee abduction moments, however, not to levels that matched the lower initial measures of the
“low risk” group. The researchers concluded that targeting those individuals with a “higher risk” may make an impact on risk of injury but that more intervention and research into intervention methods is warranted (Myer, Ford et al. 2007).

In 2008, the International Olympic committee (IOC) engaged the assistance of Renstrom, et al. (Renstrom, Ljungqvist et al. 2008) in establishing an updated current concepts statement with regards to non-contact ACL injuries in female athletes. In their work, the researchers embellished on kinematic analysis of dynamic variables relating to incidence of injury. They stated that women typically landed with less knee flexion and employed greater knee valgus throughout the stance phase. Additionally, in reviewing related intervention strategies, the research group identified the beneficial effects of proprioceptive training methods (typically as part of a multi-disciplinary, multi-faceted approach) as effective in modulating the risk of injury specifically by targeting dynamic loading of the knee joint and reducing the impulse (by increasing amortization time). These researchers also concurred that neuromuscular fatigue may directly affect biomechanics and limit proprioceptive protection of the ACL (Renstrom, Ljungqvist et al. 2008).

Krosshaug, et al. published a 2007 study that exclusively involved analysis of video segments involving game-situation ACL injuries to male and female athletes. An international panel of experts in the field each evaluated the kinematics of the mechanism of each injured player at the time of initial ground contact and at 50ms later. The results of the comparison of all reviewers’ data supported the finding that female athletes had 5.3 times higher relative risk of sustaining a valgus collapse than did male athletes in the study (Krosshaug, Nakamae et al. 2007).
In 2009, Hewett, et al. conducted a video analysis study of trunk and knee motion during non-contact ACL injuries of female athletes. In reviewing and objectively quantifying video data from 23 video samples (10 female ACL injuries, 7 male ACL injuries, and 6 female non-injured controls), the researchers were able to successfully determine lateral trunk and knee abduction angles in females as compared to males. As a result, they found that both lateral trunk and knee abduction measures were higher in ACL injured female athletes as compared with males. Additionally they noted that female athletes exhibited less forward trunk lean than female controls. As a result, the researchers concluded that females who injured their ACL exhibited greater lateral trunk motion (displacement of COG) as well as greater knee abduction, supporting current and previous research findings (Hewett, Torg et al. 2009), while adding an additional component (trunk lateral displacement) that was not previously examined to a significant degree in published literature.

2.6 Kinematics and Associated Dynamic Valgus

With the advent of, and advances in, modern 3D motion analysis and 2D video-based systems, as well as force plate systems, the area of research into the dynamic aspects relating to functional knee instability have been taken to another level entirely. A number of impactful studies have closely examined the 3D kinematics and kinetics that are at play in landing, cutting, and acceleration, particularly with respect to associated incidence or predisposition to ACL injury (Nyland, Shapiro et al. 1997, McLean, Neal et al. 1999, Malinzak, Colby et al. 2001, Ford, Myer et al. 2003, McLean, Huang et al. 2004, Ford, Myer et al. 2005b, Hewett, Myer et al. 2005, Ford, Myer et al. 2006, Sigward and Powers 2006, Myer, Ford et al. 2007, Zazulak, Hewett et al. 2007a, Borotikar, Newcomer et al. 2008, Renstrom, Ljungqvist et al. 2008, Quatman and Hewett 2009).

Such studies have also been combined with longitudinal epidemiological data to give a more astute picture of the correlation between functional dynamic knee strategies and knee injury, particularly to the ACL. As there were many dynamic kinematic studies reviewed, a highlight of some of the most prominent studies is provided herein:

McLean, et al. (McLean, Neal et al. 1999) studied the knee joint kinematics of thirty athletes (16 male and 14 female) during a side-step cutting maneuver and the relationship between that specific motion and potential for knee injury in female athletes. All subjects were evaluated with 3D motion analysis (four camera system) and force plate systems while performing a series of randomly assigned straight-line running and sidestep cutting maneuvers. Upon gathering the test data, knee joint kinematics were calculated and analyzed for comparison. The results of this study indicated that gender
differences did not contribute to differences in forward or side-step cutting biomechanics. However, the female subjects did display increased inter-trial variability for axial rotation patterns during cutting when compared with males.

Ford et al., in 2003, produced a study on valgus knee motion during landing in high school male and female basketball players. In their study, 3D kinematic analysis was employed to analyze the kinematics about the knee upon landing from a drop vertical jump. The results demonstrated a significantly higher knee valgus angle in the female players as compared with the males, thus allowing the researchers to conclude that female high school basketball players exhibited a greater average valgus knee angle upon landing which may explain the increased rate of non-contact injury experienced by female athletes (Ford, Myer et al. 2003).

McLean, et al., this time in 2004, evaluated sagittal-plane-only biomechanics with regards to the influence of isolated sagittal plane forces on ACL mechanism of injury. In this relatively small cohort study, subjects performed dynamic change of direction (single-leg side-step cut) tasks, at sport-specific speeds. All trials were evaluated utilizing 3D motion analysis (six camera system). This data was then used to build a computer model for virtual subjection to high sagittal plane forces (2000 N) in order to determine the effects of such a force. The results of this study showed that purely sagittal plane forces alone are not sufficient enough to injure the ACL (McLean, Huang et al. 2004).

In 2005, Hewett, et al. published a prospective study that evaluated the impact of biomechanical measures of neuromuscular control on risk of ACL injury in female
athletes. In their study of 205 athletes, 3D kinematics and kinetics were measured during a jump-landing task. Of particular interest in this study was the evaluation of knee abduction angles upon landing. Female athletes who injured their ACL demonstrated significantly higher (8 degrees greater) knee abduction and 20% higher ground reaction force than uninjured subjects. The research group concluded that the knee motion (in particular knee abduction in the frontal plane) and associated knee loading are predictors of ACL injury risk in female athletes (Hewett, Myer et al. 2005).

Ford, et al., in 2006, studied coronal plane excursion between male and female athletes in performing single leg landing tasks. All subjects (11 male and 11 female) performed medial and lateral drop landing tasks which were evaluated kinematically via 3D motion analysis. The results of the study showed that female athletes demonstrated decreased capacity to control lower extremity coronal plane excursion and thus were at a greater risk for non-contact ACL injury (Ford, Myer et al. 2006).

Myer, et al., in 2007, published a study that screened subjects (eighteen female athletes) for level of ACL injury risk and subsequently classified these subjects as either “high risk,” or “low risk” based on their pre-training knee abduction moments. Knee abduction and calculated moments were determined via 3D motion analysis (eight camera system), utilizing 37 retro-reflective markers. Intervention-based training was consequently conducted and all test subjects were re-evaluated after seven weeks of said training. The results of this study demonstrated significant improvement in proprioceptive neuromuscular control and dissipation of ground reaction forces as compared with pre-training data in the “high-risk” group, however, these post-test moments and GRF’s did not decrease to the same level of the “low-risk” group pre-
training values. The researchers, in conclusion, advocated further research into improvements in efficacy and efficiency of neuromuscular training programs designed to mitigate ACL injury risk in the female athlete (Myer, Ford et al. 2007).

In 2008, Borotikar, et al. released a study that focused on 3D kinematic evaluation of lower extremity joint rotations when performing landing tasks, specifically with regard to female athletes. In this study, the research team evaluated both anticipated and unanticipated single leg landings under fatigued and non-fatigued conditions. Data collected showed that fatigue had a significant effect on initial contact hip extension and on hip internal rotation as well as in peak stance knee abduction and internal rotation (Borotikar, Newcomer et al. 2008).

Zazulak, et al. (Zazulak, Hewett et al. 2007a), in an at-that-time-unique study, examined the effects of neuromuscular control of the trunk (core) on knee injury risk. In their study, lateral trunk displacement values upon unanticipated release of externally applied contra-lateral resistance were determined and plotted against risk of ACL injury. They found a significantly high sensitivity and moderate specificity with regards to the relationship between said core stability and ACL injuries in female athletes. Noteworthy is the fact that additional and, as of this day, current studies have embraced and encompassed the potential for such a causal relationship between level of trunk proprioceptive control and risk of knee injury (Zazulak, Hewett et al. 2007a).

Recently, Quatman and Hewett (Quatman and Hewett 2009) published a review of the evidence regarding non-contact ACL injury mechanisms in female athletes, specifically embellishing on the “valgus collapse” mechanism. They noted that this
“valgus collapse” does not (although the term valgus may imply so) indicate that the injury occurs solely in the frontal plane of motion. The researchers state that although much previous research has focused on either frontal or sagittal plane biomechanics, the valgus collapse mechanism is due to multi-planar factors (sagittal, frontal, and transverse). Interestingly, they state that, as a result, intervention programs that target specifically sagittal plane mechanics are less likely to be effective (Quatman and Hewett 2009).

Also in 2009, a study that has been cited previously in this review was published by Renstrom, et al (Renstrom, Ljungqvist et al. 2008). Conducted at the request of the International Olympic Committee (IOC) this study evaluated a plethora of current concept theories relating to a variety of female ACL injury related subjects from epidemiology and mechanism to surgical intervention and prevention. As part of this comprehensive study, they elaborated on the current state of biomechanics research being conducted specifically with regard to the incidence of ACL injury in the female athlete. These researchers state that results of 3D based kinematic study of ACL injury in the female athlete support lower extremity valgus (knee abduction) loading and anterior tibial translation as the likely predominant mechanism for ACL injury (Renstrom, Ljungqvist et al. 2008).

2.7 Open-NI and Kinect-Based Skeletal Tracking for Biofeedback

Recently, advances in open-source natural interaction interfaces for human gesture recognition and resultant human skeletal framework have changed the dynamics
of traditional motion analysis (Borenstein 2012, Clark, Pua et al. 2012, de Albuquerque, Moura et al. 2012, Dutta 2012, Lowes, Alfano et al. 2013, Mentiplay, Clark et al. 2013). With the advent of new hardware and software systems, the Israel-based PrimeSense has provided Microsoft (as in their XBox Kinect® system) and others with a platform capable of interfacing human movement, in real-time, with tracking algorithms capable, ultimately, of capturing and recording human kinematics with precision accuracy, outside of a traditional motion analysis laboratory (Wong 2011, Borenstein 2012).

Traditional biofeedback interventions and neuromuscular screening programs have relied on accurate, but cumbersome and extremely expensive 3D motion analysis camera arrays. The accuracy provided by such facilities has been a plus, but the major limitation is the lack of portability, high-cost, and steep learning-curve for training clinicians (Myer, Ford et al. 2004, Myer, Ford et al. 2010b).

Since the release of the Kinect system and subsequent release of open-source code for the controls, just over two years ago, many exciting and potentially revolutionary human condition benefiting projects and research studies have begun (Raptis, Kirovski et al. 2011, Borenstein 2012, Clark, Pua et al. 2012, de Albuquerque, Moura et al. 2012, Dutta 2012, Nergui M 2013).
2.8 Summary of the Literature Reviewed

Through the diligence of many talented researchers and coalitions, much has been learned over the past fifteen years about the differences in the predisposition to ACL injury and the associated mechanisms of injury in the female athlete.

What is clear from the published data is that the female knee, on average, functions differently than the male athletic knee in terms of dynamic neuromuscular recruitment and kinetic dissipation of ground reaction forces (Myer, Ford et al. 2007). It has been shown that the susceptible female knee has a greater likelihood to experience greater valgus knee motion (Bendjaballah, Shirazi-Adl et al. 1997) in both landing and in unanticipated changes of direction (McLean, Neal et al. 1999, Ford, Myer et al. 2003, Hewett, Myer et al. 2005, Ford, Myer et al. 2006, Myer, Ford et al. 2007, Zazulak, Hewett et al. 2007a, Borotikar, Newcomer et al. 2008, Renstrom, Ljungqvist et al. 2008, Quatman and Hewett 2009). This dynamic valgus (or lower leg abduction as it is sometimes referred to in the literature) is also magnified by dynamic changes in the athlete’s center of mass that often occurs with unanticipated or inefficiently controlled deviations in the translation of the core’s center of mass with relationship to the leg/foot in contact with the ground (Hewett, Myer et al. 2004, Myer, Ford et al. 2007, Borotikar, Newcomer et al. 2008). As a result of the concert between strong epidemiological, EMG, and 3D motion-analysis biomechanics research, it is clear that the female athletic knee uses different compensation strategies than the male athletic knee.

What is not clear from the published data is the extent to which independent, and notably often less easily modifiable, factors relate to predisposition to ACL injury in the
female athlete (Baker 1998, Hughes and Watkins 2006, Renstrom, Ljungqvist et al. 2008). These still somewhat unclear areas include: the influence of female hormonal changes and the onset of menstruation, and the inherent physiological characteristics (Q-angle, femoral notch width, ACL bundle tissue thickness, and generalized joint laxity) and their relationship to increased ACL injury susceptibility (Baker 1998, Hughes and Watkins 2006, Renstrom, Ljungqvist et al. 2008). Although many studies make compelling arguments for or against the levels of contributions of these aforementioned factors, there are too many conflicting studies to make sound judgment. When the issue of this confliction is considered together with the fact that many of these variables are not, or at least not efficiently, modifiable, one understands the trend to explore and expand research on the components of susceptibility that are more readily modifiable. From this collective realization has dawned the current state of research into multi-factorial models, a focus on modifiable factors, and development of associated intervention-based strategies.

2.9 Clinical Relevance

With the dramatic rise in female athletic activity and level of intensity over the past two decades (Baker 1998, White, Lee et al. 2003, Majewski, Susanne et al. 2006, Hewett 2008, Shultz 2008), so too has risen both the overall incidence and rate of female athlete ACL injury; (Baker 1998, White, Lee et al. 2003, Majewski, Susanne et al. 2006, Hewett 2008, Shultz 2008) the latter of which has led to a drive for increased and enhanced exploration into intervention-based strategies. Particularly in the higher-risk
sports of basketball and soccer, with rates in excess of five times their male counterparts (Ford, Myer et al. 2003, Agel, Arendt et al. 2005, Majewski, Susanne et al. 2006, Krosshaug, Nakamae et al. 2007), more research into sport-specific proprioceptive perturbation strategic dissipation of ground reaction forces and the associated intervention-based training regimens must be conducted.

In the modern world of advanced biomechanical analysis tools and techniques, such as 3D motion analysis and integrated force plate data, it is now possible to develop methods to decrease, although albeit unfortunately not eliminate, the susceptibility of the female athletic knee to ACL injury (Myer, Ford et al. 2004, Renstrom, Ljungqvist et al. 2008). It is the hope of this author, and that of the industry as a whole, to see the gap bridged between the rate of the female athlete and their male counterparts, therein improving the quality of the athletic experience for the individual and team, while also mitigating the economic impact on the micro and macro community.

With this said, an area of great promise in the area of ACL research, is the future application of clinic, or training facility-based (“in the field”), OpenNI / Kinect-based skeletal tracking and resultant biofeedback of real-time motion for the proactive intervention of abhorrent neuromuscular biomechanics related to non-contact ACL injury in female athletes (Clark, Pua et al. 2012, de Albuquerque, Moura et al. 2012, Dutta 2012).
Chapter 3

Methodology

3.1 Overview of Study Design

This study was of a true experimental design, consistent with the fact that intervention via manipulation of the primary independent variable (feedback) was conducted. Both within-subjects and between-subjects designs were employed where possible in an effort to ensure that control over inter-subject physiological characteristics is maintained and the intervention-related variable is isolated, as isolation of the experimental variable is necessary to eliminate selection bias and other confounding factors in a study capable of establishing causal effects of the interventions (Borman 2002).

3.2 Subjects

In this study, forty young competitive female gymnasts, between 13 and 18 years of age, who were vigorously physically active at least three days per week for at least 30 minutes per day, were initially approached. Subjects were excluded if they had a history of lower extremity musculoskeletal injury within the past three months. They were also
required to have intact bilateral ACLs, no history of lower extremity or spine orthopaedic surgery, no neurological disease, not be under the influence any neurological medications, nor had any concussions within the past three months. A small number of subjects were unable to complete the study as a result of regular practice-related ankle sprain (n=4), concussion (n=1), and those who quit the sport or team (n=3). As a result of exclusions due to the aforementioned constraints as well as a result of study drop-outs, the final main sample size was 24 subjects (Figure 3-1), each of whom were randomly assigned to either the control (CRTL) or Kinect-based feedback (KBF) group. Mean age was 14.96±1.63 years. Mean height was 1.60±0.07 meters. Mean subject mass was 53.4±7.1 kg. Mean competitive gymnastics ability level was level 7 (level is ability-based with possible minimum of 3 and maximum of 10). Of the 24 total subjects, a subset (n=8) were chosen at random to participate in the system validation portion of the study.

![Flowchart](image-url)

**Figure 3-1:** Subject Group Recruitment and Exclusion Flowchart
After the study was explained and the participants and parent/guardian had a chance to review and sign the approved University of Toledo Institutional Review Board (IRB) approved (see Appendix D for approval documentation) consent forms, all subjects completed a health and injury history questionnaire (Appendix B). Once this was complete, each participant was weighed on a standard clinical balance scale to determine accurate mass (in kilograms) as well as measured for accurate height (in meters). Upon review of the content of the questionnaire, and assuming the potential subject met all inclusion criteria, the subject was enrolled into the study.

3.3 Procedures

There were two integrated, but independently designed, components to this study: (1) design of a new real-time Kinect-based DVJ screening and biofeedback system, and (2) an intervention training effect study. These two components will be initially detailed separately below.

3.3.1 Real-time Screening and Biofeedback System Design

The first task central to this study centered around the design and engineering of a new Kinect-based biofeedback and screening system (KBBFSS). The primary components of this system included a new commercially available infrared depth-camera (PrimeSense, Tel Aviv, Israel) as depicted in Figure 3-2, open source driver software packages (OpenNI and NITE, PrimeSense, Tel Aviv, Israel), and open source
programming and graphical user interface software (Processing Foundation, processing.org) run on a Microsoft Windows (Microsoft, Redmond, WA) operating system personal computer. Similar hardware and software technology was first released commercially in the United States in 2011 as part of the Microsoft Xbox® Kinect game platform. Since that time, the system’s developer, (PrimeSense) as well as Microsoft, have released independent PC-based software development kits (SDKs) for development and to encourage future application development (Borenstein 2012). It is important to note that the infrared light spectrum that was utilized in this system is emitted at a wavelength that is invisible to the human eye and thus non-distractive to performance.

Figure 3-2: PrimeSense / Kinect Depth Camera Hardware
From the hardware perspective, in collecting an image with the PrimeSense depth camera (Figure 3-2), infrared light is actively emitted by the camera-mounted infrared projector, covering a 12 foot in depth, 58° horizontal and 45° vertical field of view from which data can detected by the adjacent CMOS image sensor and rapidly processed into a 3D “depth map” with the factory pre-calibrated PrimeSense PS1080 SoC (system on a chip) processor (Wong 2011). These data were then transmitted via USB cable to the PC-based NITE middleware software (PrimeSense, Tel Aviv, Israel) where it’s raw three-dimensional coordinate data can be interpreted by the OpenNI software at a resolution of 1024 x 758 pixels (VGA) (Wong 2011, Borenstein 2012) and utilized for collection, data storage, and feedback. This process is in some ways similar to LIDAR (laser induced differential absorption radar - a laser light based radar system) based technology, but with a shorter range and far lower cost (Wong 2011). From a precision standpoint, the system resolution is 10mm (d) x 3mm(h) x 3mm(v) (Wong 2011) at an intermediate depth (~3m) of field, which can be converted to a relative 10mm(z) x 3mm(x) x 3mm(y) traditional motion capture coordinate resolution.

For this study, a prototype portable clinical Kinect-based biofeedback and screening system (KBBSS) cart was fabricated (Figure 3-3) utilizing a rolling metal cart capable of housing the Windows 8 (Microsoft Corp., WA, USA) laptop PC (Lenovo Electronics, Beijing, China), all cables and power supplies, and a top-mounted 24 inch LED flatscreen monitor (Samsung, South Korea).
In order to achieve real-time skeleton tracking display to a PC viewing screen, the raw three-dimensional coordinate data from the camera are filtered and processed by an imbedded proprietary software algorithm wherein “skeleton” segments are fit to the subject being recorded and overlaid as visual line segments on the viewing screen (Borenstein 2012). This is achieved through software-derived pixel differentiation-based shape tracking for two dimensional images in conjunction with the third dimensional datapoint for every pixel yielded from the depth value. These segments are tracked and plotted in real-time with minimal latency at a rate of 30Hz in all three dimensions and planes of movement (Wong 2011, Borenstein 2012). As a result, frontal and sagittal plane angular rotations and displacements were calculated and displayed in real-time.

In the present study, we employed a custom-designed software application (Figure 3-4), based in a Processing (Processing Foundation, processing.org) software
application, that utilized NITE and open source algorithms and functions to track and plot knee flexion (sagittal plane), knee abduction (frontal plane) and relative knee separation distance (between left and right knee) in real-time. Additionally, the derived angles and distances were utilized to drive the visual and auditory biofeedback components. The complete program code for this study is presented in Appendix A of this document.

Figure 3-4: System Graphical User Interface (GUI)

The real-time visual biofeedback component consisted of an on-screen graphical representation of limb and joint segments that respond intuitively and immediately to subject movement within the capture volume. First, all subjects were recognized and the system calibrated by producing the system-specific calibration (“Psi”) pose (Figure 3-5) until the system successfully recognized the subject and began tracking her movements. Next, a dynamic line was anthropomorphically-fit by the system to match the subject’s intercondylar (knee) distance. As the subject’s knee separation distance increased or
decreased, the length and color of the line changed accordingly. The color of the line was displayed in one of two colors: red or green. When the distance was less than the desired threshold, the line color was displayed as red, and when the distance was equal to or greater than the threshold, the line color was displayed as green (Figures 3-6 & 3-7). The threshold was automatically subject-specifically calculated by the system, by computing the inter-ASIS distance of the subject and setting the medial knee distance threshold to fifty percent of this distance (Noyes, Barber-Westin et al. 2005). It is important to note that we chose to utilize this subject-specific scaling as a method of normalization to subject height (relative) rather than by absolute value.

The real-time magnitude of the distance vector was displayed in millimeters at the top of the screen and the color was set to match the configuration aforementioned. Additionally, with regards to knee flexion angle, bilateral knee flexion angles were calculated in real-time and displayed, in units of degrees, at the top of the screen. This text was displayed in green when the value was greater than the knee flexion threshold and in red when the value was less than the set threshold (Figures 3-6 & 3-7).
Figure 3-5: System Initial Subject Recognition and Calibration (‘Psi’) Pose

Figure 3-6: ‘Red’ Line/Text On-Screen Feedback
The auditory real-time biofeedback component was tied directly to the knee medial intercondylar distance value as discussed in detail previously. When the distance was at, or above, the calculated threshold, there was no auditory feedback. However, when the distance dropped below the threshold, the system emitted, via the PC speakers, an audible tone indicative of the threshold breach.
3.3.2 Biofeedback Intervention Procedures

All participants who met the inclusion criteria were introduced to a 30cm drop vertical jump landing task, consistent with the procedure previously used by many ACL screening studies (Hewett, Myer et al. 2005, Myer, Ford et al. 2007, Hewett, Torg et al. 2009, Myer, Ford et al. 2010b, Myer, Ford et al. 2011). Additionally, a live demonstration was performed for them by a member of the research team. Next, they had the opportunity to practice the task five times. Once the subject was comfortable performing the task, initial kinematics (the “pre-test”) were collected non-invasively using a single PrimeSense (PrimeSense, Tel Aviv, Israel) infrared depth camera system centered directly in-front of the subject at a distance of 3 meters and a height of 1 meter relative to her jump landing area (as depicted in Figure 3-8). This set-up allowed simultaneous recording of peak knee flexion angle in the sagittal plane as well as peak knee separation distance in the frontal plane.
All subjects were then randomly assigned into one of two groups via a Microsoft Excel-based random number generator. The two primary classifications were: an intervention group (KBF) that performed drop vertical jump landing training while using the KBBFSS and a control group (CTRL) that performed the same task but without KBBFSS feedback.

Additionally, upon later review of the kinematics collected for each subject, we utilized the data to stratify the participants as either: “high risk” or “normal risk” for non-contact ACL-injury by utilizing a novel risk stratification algorithm outlined below.
Specifically, this was done by plotting each subject’s relative average frontal plane knee valgus angle with sagittal plane knee flexion angle, for seven frames before and after the frame representing initial ground contact (Figure 3-9) wherein the slope of the best fit line for each was calculated (Figure 3-10) and converted to a relative angle in degrees utilizing an arctangent function in Microsoft Excel (Microsoft Corp, WA, USA). The resultant net difference between these two angles was then calculated. A negative net value was indicative of a “convergence” of the two slopes while a positive net value indicated a parallel or “divergence” of the two slopes. Those subjects with a net negative resultant value were stratified as “high risk” and those with a net positive value were stratified as “normal risk”.
Figures 3-9A&B: Slope of Each Key Variable: Pre & Post (Sample Subject)
Upon assignment to a group, all subjects began the study immediately. For the Kinect-based biofeedback group the study intervention was as follows: Each subject performed a series of twenty 30cm drop landings using the infrared video-based real-time biofeedback tool three days per week for four consecutive weeks. Subjects were instructed to complete each jump separately, using the images presented to them on 24 inch flat-screen LED monitor (Samsung Electronics, Suwon, South Korea) placed at average subject eye-level (48 inches from floor) and 60 inches in front of the landing zone, to augment control of their landing mechanics (Figure 3-11). The display presented simple graphical (lines) and a quantitative (numerical data) representations of their kinematics as well as auditory indications as described in the previous section. A green line connecting the knees indicated proper spacing in the frontal plane, while a red line indicated aberrant mechanics. Additionally, knee flexion angle was presented in accordance with the same feedback methods. All feedback was presented in real-time. These sessions were monitored by a member of the research team and/or coaching staff to

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**RISK QUANTIFICATION STRATIFICATION ALGORITHM SLOPE FORMULA**

\[
\text{SLOPE 1} = \text{DEGREES (ARCTAN (SLOPE (POINTS ON LINE 1, HORIZONTAL AXIS)))}
\]

\[
- \quad \text{SLOPE 2} = \text{DEGREES (ARCTAN (SLOPE (POINTS ON LINE 2, HORIZONTAL AXIS)))}
\]

\[
= \quad \text{NET RESULT}
\]

If the net result of the difference between the slopes is:

- POSITIVE \(\rightarrow\) “NORMAL” RISK
- NEGATIVE \(\rightarrow\) “HIGH” RISK

Figure 3-10: RQS Algorithm Slope-Based Calculation
insure that subjects performed the tasks as outlined and without any additional commentary or feedback beyond the standard initial script outlined below.

The control group performed the same activity at the same frequency and intensity, but without the KBBFSS (no visual nor auditory feedback). Control subjects were all read the same statement prior to training sessions:

“Please perform all jumps with a focus on soft landings, remembering to bend your knees and keep your knees out rather than allowing them to buckle towards each other during landing. A soft and quiet landing should be attempted.”

Figure 3-11: Subject Utilizing KBF during DVJ
Upon completion of the four week intervention period, all participants were re-tested utilizing the same methods for collection of kinematics used during the pre-test component.

We also randomly selected a representative sample from the main cohort of 24 (n=8) subjects for additional testing utilizing the University of Toledo Motion Analysis Lab’s 3D motion capture (MOCAP) system in order to perform a validation and reliability comparison of the two systems. The kinematic variables assessed for reliability and validity included: knee separation (functionally representative of knee valgus), and knee flexion angle. Additionally, we collected vertical ground reaction forces (GRF) and knee valgus moment as additional kinetic measurements for those subjects who were allocated to this sample subset.

3.3.3 Lab-Based 3D Motion Capture Instrumentation and Procedures

During the 3D motion capture sessions, each subject was instrumented with a standard thirty-six 20mm spherical retro-reflective lower extremity marker set (Figures 3-14, 3-15, and 3-16), enabling 3D kinematic data to be captured at 200hz using Cortex version 3.3.1 motion analysis software (Motion Analysis Corp, CA, USA) via a 12-unit Eagle IR camera array (Motion Analysis Corp, CA, USA). Two independent 50cm x 50cm AMTI model OR-6-5 AMTI force plates (AMTI, MA, USA) sampled ground reaction forces at 2000Hz as well as well as facilitated infrared camera matched-inclusion of this data into the Cortex analysis software. The PrimeSense infrared camera collected data using the same set-up to capture data as was used in the clinical environment.
component of the study and concurrent with the 3D MOCAP system in order to enable system validation assessment. Figures 3-12 and 3-13 depict the lab set-up configuration.

Figure 3-12: Lab Configuration Denoting Camera & Force Plate Locations
Figure 3-13: Lab Configuration Denoting Camera & Force Plate Locations
Figure 3-14: Subject Marker-set Configuration (Anterior)

Figure 3-15: Subject Marker-set Configuration (Posterior)
With both systems collecting simultaneously, a one-second standing baseline static trial was captured, per industry-standardized protocol, with the subject standing with one foot on each force-plate and arms held out to the side. Subsequently, up to five task specific practice trials were performed from the 30 cm box to the landing area. Then, five successful 5-second trials of the 30cm drop vertical jump landing task were captured and the kinematic and kinetic data stored for analysis.

Figure 3-16: 3D MOCAP Model Depicting Marker Locations
3.4 Data Processing

All pre-training and post-training KBBFSS-derived raw data were exported to Microsoft Excel (Microsoft Corporation, USA) where they were smoothed utilizing a 9-point moving average window. Peak knee flexion and knee separation distance values were then calculated and averaged for each subject across trials as well as for each training group. All MOCAP kinematic data was initially processed using Cortex (Motion Analysis Corporation, CA, USA), then exported to Visual 3D (C-Motion, MD, USA) analysis software where the data was low-pass filtered at 12 Hz and a dynamic lower-body 3D model (Figures 3-15 and 3-16) was applied. Kinematics and inverse dynamics were calculated in Visual 3D. Ground reaction forces (Fx, Fy, Fz), knee joint angles (frontal and sagittal plane) and knee joint moments (frontal plane) were calculated for each subject for the timespan ranging from initial ground contact to 30 ms after initial ground contact.

For the validation component, MOCAP and clinical system data derived peak values were matched and trended statistically as outlined in the following section.
3.5 Statistical Analysis

All statistical analyses were performed using SPSS version 17.0 (IBM, NY, USA). For the validation component of this study, Pearson’s correlations were calculated for assessment of the concurrent validity of the systems. Intersession reliability was explored with interclass correlations (ICC$_{2,1}$). For the training intervention component, a $2 \times 2$ mixed analysis of variance (ANOVA) was performed first to test for main effects and significant interactions. Follow-up Bonferroni post hoc tests explored statistically significant main effect findings. Finally, for the RQS component, ANCOVAs followed by linear regression analysis for significant findings was performed. The alpha-level was set a priori at $p \leq 0.05$ for all statistical comparisons.
3.6 Internal Validity

Internal validity, as outlined by Onwuegbuzie (Onwuegbuzie 2003), may have been potentially limited in this study, specifically with regards to instrumentation, as kinematic data variability may have been related to marker locations and movement artifact, which could have yielded additional noise in the data. All reasonable efforts were taken to constrain these locations and site preparation parameters in order to mitigate this error through the use of industry standardized proper skin preparation procedures. Also, the differential selection of participants may have been compromised by selecting subjects only from the local athletic community, and all from the same sport (gymnastics).
Chapter 4

Results

4.1 Demographics and Baseline Data

A complete demographic and anthropometric data table for all subjects is presented below (Table 4.1). All subjects were female. Mean age was 14.96±1.63 years. Mean height was 1.60±0.07 meters. Mean subject mass was 53.4±7.1 kilograms. Mean competitive gymnastics level was Level 7.

While there was a significant difference in subject age (p=0.03) between groups, the difference was pragmatically small (mean age difference: 1.3±1.4 years). Importantly, there was no significant difference in subject competitive gymnastics ability level (p=0.664). There were also no significant differences in subject height (p=0.229) or mass (p=0.211) anthropomorphic characteristics between the control DVJ training group and the KBF training group (Table 4.1). Additionally, there were no significant differences in the subject characteristics between the main cohort of 24 subjects and that of the subset of 8 subjects randomly chosen to represent the system validation sample. For a complete table of subject anthropomorphic and all results, see Table C.1 in Appendix C.
 Appropriately, there were no significant differences between groups in pre-training average knee flexion \((p=0.146)\), nor for knee separation \((p=0.855)\) values. Finally, it is also important to note that there was no significant difference in the mean number of training sessions completed \((p=0.905)\) between CTRL and KBF groups (Tables 4.2 & 4.3).

Table 4.1: Subject Anthropomorphic and Demographic Data

<table>
<thead>
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<th>Subject</th>
<th>Height</th>
<th>Mass</th>
<th>Age</th>
<th>Sex</th>
<th>Group</th>
</tr>
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<td>#</td>
<td>in</td>
<td>m</td>
<td>lb</td>
<td>kg</td>
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<td>578.22</td>
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<tr>
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Table 4.2: Group Statistics for Number of Intervention Training Sessions Completed

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<th>Std. Deviation</th>
<th>Std. Error Mean</th>
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<td></td>
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<td>.52884</td>
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Table 4.3: Independent Samples Test for Intervention Training Sessions Completed

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<th>Sessions Completed</th>
<th>Equal variances assumed</th>
<th>t</th>
<th>df</th>
<th>Sig. (2-tailed)</th>
<th>Mean Difference</th>
<th>Std. Error Difference</th>
<th>Lower</th>
<th>Upper</th>
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<td>.68856</td>
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</tr>
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</table>

Table 4.4: Group Mean and SD for Age (yrs), Height (m), and Mass (kg)

<table>
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<th>Std. Deviation</th>
<th>Std. Error Mean</th>
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<td>Control</td>
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<td></td>
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Table 4.5: Independent Samples t Test for Group Differences in Age

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<tr>
<th>t</th>
<th>df</th>
<th>Sig. (2-tailed)</th>
<th>Mean Difference</th>
<th>Std. Error Difference</th>
<th>95% Confidence Interval of the Difference</th>
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<td>.030</td>
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<td>-2.67914, -.15419</td>
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Table 4.6: Independent Samples t Test for Group Differences in Subject Height

<table>
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<tr>
<th>t</th>
<th>df</th>
<th>Sig. (2-tailed)</th>
<th>Mean Difference</th>
<th>Std. Error Difference</th>
<th>95% Confidence Interval of the Difference</th>
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</thead>
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<td>-1.24</td>
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<td>.229</td>
<td>-.035083</td>
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<td>-.093823, .023656</td>
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</tbody>
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Table 4.7: Independent Samples t Test for Group Differences in Subject Mass

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<td>t</td>
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<tr>
<td></td>
<td></td>
</tr>
<tr>
<td>Weight Equal variances assumed</td>
<td>-1.29</td>
</tr>
</tbody>
</table>
4.2 Validity and Reliability

4.2.1 Validity

When evaluating knee flexion and knee separation tracking peak kinematic values for 30ms following subject initial ground contact, the KBBFSS and the MOCAP system yielded valid, highly matched results (Table 4.8). Pearson product moment correlations produced an r of 0.963 (p< 0.001) for pre-training knee flexion and an r of 0.897 (p=0.001) for post-training knee flexion. For knee separation, Pearson product moment correlations yielded an r of 0.815 (p=0.007) for pre-training and an r of 0.916 (p< 0.001) for post-training.

Table 4.8: System Comparison Pearson Correlations for Knee Flexion and Separation

<table>
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<th>Knee Separation (Frontal Plane)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Pre</td>
<td>Post</td>
</tr>
<tr>
<td></td>
<td>Kinect v MOCAP</td>
<td>Kinect v MOCAP</td>
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<tr>
<td>Pearson r</td>
<td>0.963</td>
<td>0.897</td>
</tr>
<tr>
<td>Significance (p&lt;.05)</td>
<td>p=0.001**</td>
<td>p=0.001**</td>
</tr>
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</table>
4.2.2 Reliability

*Please note that the following reliability results were obtained from subjects who were all undergoing an intervention training study. As a result, changes in peak kinematics were expected. The results should be evaluated with caution as the intent was to compare systems rather than reliability between sessions:*

When evaluating knee flexion and knee separation tracking peak kinematic values for 30ms following subject initial ground contact, the KBBFSS and the MOCAP system were comparably reliable (Table 4.9), particularly for knee separation distance. For the MOCAP system, the inter-session (pre to post testing session) ICC\(_{2,1}\) was 0.906 (p=0.003) and 0.701 (p=0.067), for knee separation distance and knee flexion angle, respectively. For the KBBFSS, the inter-session (pre to post testing session) ICC\(_{2,1}\) was 0.853 (p=.011) and 0.471 (p=0.210), for knee separation distance and knee flexion angle, respectively.

Table 4.9: ICC for Inter-Trial Reliability of KBBFSS and MOCAP Systems

<table>
<thead>
<tr>
<th></th>
<th>Knee Flexion (Sagittal Plane)</th>
<th>Knee Separation (Frontal Plane)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>MOCAP</td>
<td>Kinect</td>
</tr>
<tr>
<td>ICC(_{2,1})</td>
<td>0.701</td>
<td>0.471</td>
</tr>
<tr>
<td>Significance</td>
<td>p=0.067</td>
<td>p=0.210</td>
</tr>
</tbody>
</table>
4.3 Intervention Training Kinematics

4.3.1 Baseline Kinematics

It is important, first, to note herein that dynamically assessed pre-training knee flexion angle (p=0.146) and intercondylar knee distance (p=0.855) values were similar between groups (Tables 4.10 to 4.13), in advance of any intervention, with no significant differences noted. In addition, there was no significant difference (p=0.905) in the number of training sessions completed, between groups (Refer to Tables 4.2 and 4.3).

Table 4.10: Group Comparison of Pre-Training Knee Flexion Angle (degrees)

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Mean</th>
<th>Std. Deviation</th>
<th>Std. Error Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pre-Training Knee Flexion Control</td>
<td>12</td>
<td>15.1667</td>
<td>4.40729</td>
<td>1.27228</td>
</tr>
<tr>
<td>Pre-Training Knee Flexion Biofeedback</td>
<td>12</td>
<td>12.5000</td>
<td>4.25334</td>
<td>1.22783</td>
</tr>
</tbody>
</table>

Table 4.11: Independent Samples Test of Pre-Training Knee Flexion Angle (degrees)

<table>
<thead>
<tr>
<th></th>
<th>t</th>
<th>df</th>
<th>Sig. (2-tailed)</th>
<th>Mean Difference</th>
<th>Std. Error Difference</th>
<th>95% Confidence Interval of the Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pre-Training Knee Flexion Equal variances assumed</td>
<td>1.508</td>
<td>22</td>
<td>.146</td>
<td>2.66667</td>
<td>1.76812</td>
<td>-1.000 6.3335</td>
</tr>
</tbody>
</table>
Table 4.12: Group Comparisons for Pre-Training Inter-Knee Distance (mm)

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Mean</th>
<th>Std. Deviation</th>
<th>Std. Error Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pre-Training Distance</td>
<td>12</td>
<td>243.8333</td>
<td>43.02818</td>
<td>12.42117</td>
</tr>
<tr>
<td>Control</td>
<td>12</td>
<td>246.6667</td>
<td>30.97898</td>
<td>8.94286</td>
</tr>
</tbody>
</table>

Table 4.13: Independent Samples Test for Pre-Training Inter-Knee Distance

<table>
<thead>
<tr>
<th></th>
<th>t</th>
<th>df</th>
<th>Sig. (2-tailed)</th>
<th>Mean Difference</th>
<th>Std. Error Difference</th>
<th>95% Confidence Interval of the Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pre-Training Distance</td>
<td>-.19</td>
<td>22</td>
<td>.855</td>
<td>-2.83333</td>
<td>15.30556</td>
<td>-34.57512 to 28.90845</td>
</tr>
<tr>
<td>Equal variances</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

4.3.2 Knee Flexion: Group x Time Comparisons

There was a significant difference in change in knee flexion angle (p=0.001) for the KBF training group (Figure 4-3) as compared with the control training group (Figure 4-2). The effect size of this statistic was large (d= 1.618, r =0.6289) and yielded a resultant power of 0.93 at the sample size of 24 subjects. Knee flexion increased 46% (p=0.002) as a result of the Kinect-based biofeedback training program (Figure 4-1), but showed no change (p=0.614) in the control group (Figure 4-2).
Figure 4-1: KBF Peak Knee Flexion Angles for Pre and Post Training

Figure 4-2: CTRL Peak Knee Flexion Angles for Pre and Post Training
4.3.3 Knee Separation Distance: Group x Time Comparisons

There was a significant difference in knee separation ($p=.024$) for the KBF training group (Figure 4-6) as compared with the control training group (Figure 4-5). The effect size of this statistic was moderately large ($d=0.99$, $r=0.44$) and yielded a resultant power of 0.73 at the sample size of 24 subjects. In sum, knee separation increased 21% ($p<0.001$) as a result of the Kinect-based biofeedback training program (Figure 4-4).
**Figure 4-4: KBF Normalized Peak Knee Separation, Pre and Post Training**

**Figure 4-5: CTRL Normalized Peak Knee Separation, Pre and Post Training**
4.4 Intervention Training: Kinetics

4.4.1 Peak Vertical Ground Reaction Force (vGRF)

While subjects in the KBF training group did not significantly reduce peak vertical ground reaction force more than the control group (p=0.450), there was a noticeable trend toward a difference and the effect size was large (Cohen’s d=1.84) despite the small subset sample size of 8 subjects (Table 4.14 & 4.15, and Figure 4-7). The KBF group reduced peak vGRF by 1.5%, whereas the CTRL group increased peak vGRF by 4.7%.
Table 4.14: Group Change in Peak Normalized vGRF (BW)s

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Mean</th>
<th>Std. Deviation</th>
<th>Std. Error Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>Control</td>
<td>5</td>
<td>.084000</td>
<td>.1653179</td>
<td>.0739324</td>
</tr>
<tr>
<td>Biofeedback</td>
<td>3</td>
<td>-.020000</td>
<td>.1967232</td>
<td>.1135782</td>
</tr>
</tbody>
</table>

Table 4.15 Independent t Test for Change in Peak vGRF

<table>
<thead>
<tr>
<th>vGRF Change Score</th>
<th>t</th>
<th>df</th>
<th>Sig. (2-tailed)</th>
<th>Mean Difference</th>
<th>Std. Error Difference</th>
<th>95% Confidence Interval of the Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>vGRF Change Score</td>
<td>.807</td>
<td>6</td>
<td>.450</td>
<td>.1040000</td>
<td>.1288306</td>
<td>-.2112372, .4192372</td>
</tr>
</tbody>
</table>
4.4.2 Peak Bilateral Frontal Plane Valgus Knee Moment

While subjects in the KBF training group did not significantly reduce peak bilateral frontal plane valgus moment more than the control group (p=0.697), there was a noticeable trend toward a difference and the effect size was moderate (Cohen’s d=0.44) despite the small subset sample size of 8 subjects (Tables 4.16-4.17 and Figure 4-8). The KBF group decreased their peak net bilateral frontal plane valgus joint moment 26% while the CTRL group decreased the same kinetic measure by 17%.
### Table 4.1: Group Change in Bilateral Peak Valgus Moment (Net Joint Moment)

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Mean</th>
<th>Std. Deviation</th>
<th>Std. Error Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>Valgus Moment Change Score</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Control</td>
<td>4</td>
<td>-.112175</td>
<td>.1220287</td>
<td>.0610144</td>
</tr>
<tr>
<td>Biofeedback</td>
<td>3</td>
<td>-.165200</td>
<td>.2195684</td>
<td>.1267679</td>
</tr>
</tbody>
</table>

### Table 4.17 Independent t Test for Change in Bilateral Peak Joint Moment (Valgus)

<table>
<thead>
<tr>
<th>Valgus Moment Change Score</th>
<th>t</th>
<th>df</th>
<th>Sig. (2-tailed)</th>
<th>Mean Difference</th>
<th>Std. Error Difference</th>
<th>95% Confidence Interval of the Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Equal variances assumed</td>
<td>.413</td>
<td>5</td>
<td>.697</td>
<td>.0530250</td>
<td>.1283001</td>
<td>-.2767809</td>
</tr>
</tbody>
</table>
Figure 4-8: Group Comparison for Change in Bilateral Peak Valgus Moment
4.5 Risk Quantification Stratification Algorithm

4.5.1 Pre and Post RQS: All Risk Levels

The correlation between the risk quantification score (RQS), based on the pre-test Kinect-based kinematic evaluation mathematical algorithm aforementioned in this study, and the DVJ training intervention resultant change in knee flexion for all risk levels ($r=0.296$, $p=0.080$), is depicted in Table 4.18.

Table 4.18: RQS and Knee Flexion Change Score – All Risk Levels

<table>
<thead>
<tr>
<th>RQS (Pre)</th>
<th>Pearson Correlation</th>
<th>RQS(Pre)</th>
<th>Knee Flexion Change Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sig. (1-tailed)</td>
<td>1</td>
<td>-.296</td>
<td></td>
</tr>
<tr>
<td>Sum of Squares and Cross-products</td>
<td>97477.871</td>
<td>-2571.675</td>
<td></td>
</tr>
<tr>
<td>Covariance</td>
<td>4238.168</td>
<td>-111.812</td>
<td></td>
</tr>
<tr>
<td>N</td>
<td>24</td>
<td>24</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Knee Flexion Change Score</th>
<th>Pearson Correlation</th>
<th>Knee Flexion Change Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sig. (1-tailed)</td>
<td>-.296</td>
<td>1</td>
</tr>
<tr>
<td>Sum of Squares and Cross-products</td>
<td>-2571.675</td>
<td>774.500</td>
</tr>
<tr>
<td>Covariance</td>
<td>-111.812</td>
<td>33.674</td>
</tr>
<tr>
<td>N</td>
<td>24</td>
<td>24</td>
</tr>
</tbody>
</table>
There was a moderate positive correlation between the risk quantification score (RQS), based on the pre-test Kinect-based kinematic evaluation mathematical algorithm aforementioned in this study, and the DVJ training intervention resultant change in knee separation distance for all risk levels ($r=0.406$, $p=0.024$) as depicted in Table 4.19.

Table 4.19: RQS and Knee Separation Change Score – All Risk Levels

<table>
<thead>
<tr>
<th></th>
<th>RQS(Pre)</th>
<th>Norm Knee Distance Change Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>RQS(Pre)</td>
<td>Pearson Correlation</td>
<td>-0.406*</td>
</tr>
<tr>
<td></td>
<td>Sig. (1-tailed)</td>
<td>0.024</td>
</tr>
<tr>
<td></td>
<td>Sum of Squares and Cross-products</td>
<td>97477.871</td>
</tr>
<tr>
<td></td>
<td>Covariance</td>
<td>4238.168</td>
</tr>
<tr>
<td></td>
<td>N</td>
<td>24</td>
</tr>
<tr>
<td>Normalized Knee Distance Change Score</td>
<td>Pearson Correlation</td>
<td>-0.406*</td>
</tr>
<tr>
<td></td>
<td>Sig. (1-tailed)</td>
<td>0.024</td>
</tr>
<tr>
<td></td>
<td>Sum of Squares and Cross-products</td>
<td>-13797.218</td>
</tr>
<tr>
<td></td>
<td>Covariance</td>
<td>-599.879</td>
</tr>
<tr>
<td></td>
<td>N</td>
<td>24</td>
</tr>
</tbody>
</table>

*Correlation is significant at the 0.05 level (1-tailed).
4.5.2 RQS and Knee Flexion: High Risk Subjects

There was a very strong positive correlation (Table 4.20 and Figure 4-9) between the risk quantification score (RQS), based on the pre-test Kinect-based kinematic evaluation mathematical algorithm aforementioned in this study, and the DVJ training intervention resultant change in knee flexion for “high risk” subjects (r=0.750, p=0.016).

Table 4.20: RQS and Knee Flexion Change Score – “High Risk” Subjects

<table>
<thead>
<tr>
<th></th>
<th>RQS(Pre)</th>
<th>Knee Flexion Change Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>RQS(Pre)</td>
<td>Pearson Correlation</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Sig. (1-tailed)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Sum of Squares and Cross-products</td>
<td>54829.426</td>
</tr>
<tr>
<td></td>
<td>Covariance</td>
<td>7832.775</td>
</tr>
<tr>
<td></td>
<td>N</td>
<td>8</td>
</tr>
</tbody>
</table>

| Knee Flexion Change Score | Pearson Correlation | -.750* | 1 |
|                          | Sig. (1-tailed)    | .016   |   |
|                          | Sum of Squares and Cross-products | -2565.010 | 213.500 |
|                          | Covariance        | -366.430 | 30.500 |
|                          | N                | 8     | 8     |

*Correlation is significant at the 0.05 level (1-tailed).
Figure 4-9: Regression for Knee Flex Change with RQS (Pre)
4.5.3 RQS and Knee Separation Distance: High Risk Subjects

There was also a strong positive correlation (Table 4.21 and Figure 4-10) between the value of the risk quantification score (RQS-Pre), based on the pre-test Kinect-based kinematic evaluation mathematical algorithm aforementioned in this study, and change in knee separation distance for “high risk” subjects ($r=0.620$, $p=0.05$).

Table 4.21: RQS and Knee Separation Change Score – “High Risk” Subjects

<table>
<thead>
<tr>
<th></th>
<th>RQS(Pre)</th>
<th>Normalized Knee Distance Change Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>RQS(Pre)</td>
<td>Pearson Correlation</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Sig. (1-tailed)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Sum of Squares and Cross-products</td>
<td>54829.426</td>
</tr>
<tr>
<td></td>
<td>Covariance</td>
<td>7832.775</td>
</tr>
<tr>
<td></td>
<td>N</td>
<td>8</td>
</tr>
<tr>
<td>Normalized Knee Distance Change Score</td>
<td>Pearson Correlation</td>
<td>-.619</td>
</tr>
<tr>
<td></td>
<td>Sig. (1-tailed)</td>
<td>.05</td>
</tr>
<tr>
<td></td>
<td>Sum of Squares and Cross-products</td>
<td>-9620.860</td>
</tr>
<tr>
<td></td>
<td>Covariance</td>
<td>-1374.409</td>
</tr>
<tr>
<td></td>
<td>N</td>
<td>8</td>
</tr>
</tbody>
</table>
Figure 4-10: Regression for Knee Distance Change with RQS (Pre)
Chapter 5

Discussion

5.1 Overview


This study aimed to accomplish two primary tasks. The first was to successfully design and deploy a clinically deployable tool capable of accurately measuring DVJ knee mechanics consistent with increased risk of non-contact ACL injury as well as to provide real-time feedback to said populations. The second task was to explore the efficacy of such an intervention tool when deployed to an environment outside of a traditional
laboratory. In sum, this study was indeed successful in providing evidence to support effective and efficient application of such a clinically-based system for both measurement and intervention utilization.

5.2 Validity and Reliability

As supported by the significant correlations denoted in the above results section, the KBBFSS produced valid kinematic values as compared with a “gold-standard” 3D motion-capture (MOCAP) lab system. These correlations are consistent with previous similar analyses in the literature (Clark, Pua et al. 2012, de Albuquerque, Moura et al. 2012). When each of the examined metrics are evaluated independently, one can see that there was indeed an accuracy bias with the MOCAP system providing more accurate knee flexion angle and knee separation distances. Additionally, the KBBFSS system more accurately measured knee separation distance than knee flexion angle. It is likely that the bias towards accuracy in the frontal plane is due to camera position. In the frontal plane, clothing was in no way a detriment as the subjects’ knees were bare and the body composition of the subjects was extremely lean, thus lending to very little artifact in the data. In the sagittal plane, as a result of the perpendicular arrangement of the camera to the sagittal plane, there was likely some error due to clothing and camera alignment not seen in the frontal plane.

The validity of the kinematics derived by the KBBFSS are of significant clinical value as this supports the application of the system to environments where traditional laboratory-based motion analysis systems were previously not financially or logistically
practical (Khoshelham and Elberink 2012). It may, in fact, now be feasible to implement objective screening and intervention measures that previously were not available to most athletes, clients, or patients. With the average capital expenditure for a fully-equipped motion capture system requiring substantial financial resources, in conjunction with additional investments required for training and maintenance, alternative systems are increasingly particularly enticing (Clark, Pua et al. 2012). Other advantages include the non-invasive and non-cumbersome nature of the markerless depth-camera approach as this can both save the clinician time and capital in accurately placing markers on each subject as well as reduce user error related to marker positioning.

5.3 Intervention Training: Kinematics

The Kinect-based biofeedback training intervention significantly improved the previously established non-contact ACL injury risk kinematics at the knee (Hewett, Myer et al. 2006a, Shultz 2008, Quatman and Hewett 2009). Specifically, the group that received Kinect-based biofeedback DVJ training gained significantly greater knee flexion and greater knee separation values, following four weeks of training, as compared with the traditional DVJ training (control) group. A sample representation, derived from the values of a single subject in the KBF group of this study, is depicted in Figures 5-1 through 5-3.

As has been well established in the literature, both excessive frontal plane peak dynamic knee valgus angle as well as decreased sagittal plane peak knee flexion angle are
suspect in the higher rate of non-contact ACL injury in the female adolescent athlete (Hewett, Myer et al. 2006a, Shultz 2008, Quatman and Hewett 2009). This combination of kinematic factors puts the ACL in a structurally compromised stretched position while also biomechanically limiting the ability of the hamstrings to effectively limit anterior translation of the tibia. While a number of studies have sought to modify these factors, via a myriad of training solutions, this particular study is one of the first to implement accurate and effective real-time visual and auditory biofeedback, specifically designed to change kinematics, in a remote (clinic or training facility) environment.

In that previous literature has pointed to numerous logistical factors that prevent effective deployment and compliance with targeted neuromuscular interventions for reducing risk of non-contact ACL injury (Myer, Ford et al. 2010a, Joy, Taylor et al. 2013), the efficacy and efficiency of the methods outlined herein may assist clinicians in improving athlete, team, and coach or administrative staff compliance with such a program.
Figure 5-1: Model of KBF Subject’s Change in Peak Knee Flexion

Figure 5-2: Model of KBF Subject’s Change in Normalized Knee Separation

Figure 5-3: Model of KBF Subject’s Peak Knee Kinematics & vGRF
5.4 Intervention Training: Kinetics

The Kinect-based biofeedback intervention showed a positive trend for decreasing aberrant non-contact ACL injury-implicated DVJ kinetics. Specifically, the KBF group showed decreased peak vertical ground reaction forces and decreased frontal plane dynamic valgus moment, following four weeks of training as compared with the control group. It is believed that excessive vGRF forces and frontal plane valgus knee joint moment are both implicated in increased non-contact ACL injury risk. Though these findings were not deemed statistically significant, it is important to note that the kinetic component utilized a subset sample of eight of the 24 subjects and thus was underpowered to provide statistical evidence herein. It should be noted, however, that because the resultant effect sizes were high and moderate respectively, there is evidence to support a potential positive interaction between the training intervention and pertinent kinetic measures.

5.5 Risk Stratification Algorithm

Our mathematically-based quantitative risk-stratification algorithm did predict outcome components for the training intervention. Specifically, subjects who were stratified as “high-risk” showed greater increases in knee flexion angle and knee separation distance. Statistical evidence herein supports currently our postulations, as well as those in previously established circles (Myer, Ford et al. 2007, Myer, Ford et al. 2010b, Myer, Ford et al. 2011), that those individuals who are at greatest functional
biomechanical risk of non-contact ACL injury stand to gain the greatest benefit from such an intervention.

Additionally, it can be further theorized that time, resource, and financial efficiency can be maximized utilizing such stratification measures. Especially in environments where resources and time are limited, resultantly derived “strategically targeted” intervention measures can ensure that the most financially and logistically efficient impact is achieved. It is, however, important to note that, although potentially promising, this analysis was calculated with a small sample size, and thus these results must be considered conservatively.

5.6 Limitations

Although appropriate measures were employed to minimize limitations to this study, some inherent limitations were present. These included sample size, homogeneity and convenience-factors of the sample, and data collection noise variables.

Sample size limitations were largely overcome for most outcome measures and statistics within this study as adequate power was reached despite same size for all but the kinetic variable component wherein the sample size was only eight subjects.

Though the sample population targeted for this study was quite homogenous considering that only female competitive gymnasts between the ages of 13 and 18 years were utilized, the benefits of such targeting outweighed the drawbacks. In this case, homogeneity afforded the capability to limit extraneous training and environmental
variables that might otherwise confound the intervention. In addition, the ability to use athletes who train and perform at all times without footwear eliminated an additional potentially confounding variable of shoe-type that would have otherwise had to be controlled for or considered.

With regards to physical 3D MOCAP marker-based limitations, it should be noted that while the markers were small (20mm) and very lightweight, some very small movement of the markers due to their inherent inertial properties may have caused residual levels of noise.

Additionally, it must be noted that attrition is always a factor and did lead ultimately to a reduction in total sample size. Some subjects were unable to complete the study due to illness, injury occurring outside of the study environment, or leaving the team. These factors are common in any study and did not statistically impact the outcome.

5.7 Directions for Future Study

While this study indicated that a KBBFSS can be effective for both screening for, and providing real-time intervention of, mechanics consistent with increased risk of non-contact ACL injury in adolescent female athletes, further research is warranted in a number of areas. First, since the population was limited to female gymnasts, the question remains as to whether these effects are transferrable to all female athletes, including those in even greater-risk sports such as soccer and basketball.
Additionally, what is the transferability of the findings to male athletes with regards to non-contact ACL injuries as it has been well established that they tend to tear their ACL in a very different manner, biomechanically?

Also, one must inquire for how long do the benefits of the intervention last in a particular individual? Though we saw significant improvements in biomechanics consistent with decreased risk of injury, how long are these improvements maintained once the real-time feedback measures are withdrawn from the training environment?

Finally, as this study measured mechanics, but not specifically injury epidemiology from an episodic standpoint, what is the impact on the rate of actual acute ACL tear rates in this, and other, subject populations? This would certainly require longitudinal study of a sufficient cohort size over an extended length of time.

5.8 Conclusion

In conclusion, this study has provided evidence in support of the accuracy and effectiveness of a clinically-deployable real-time screening and biofeedback system. Subjects who utilized the KBBFSS realized greater improvements in kinematics traditionally established as consistent with increased risk of non-contact acute ACL injury. Additionally, the KBBFSS was found to be both valid and reliable as compared to a traditional laboratory-based system (MOCAP) at a fraction of the cost and time, therein making it both efficient and effective for clinical deployment.
While additional research into the longitudinal effects on ultimate injury epidemiology as well as the transferability of these findings to alternate subject populations is warranted, this study marks an evolution in clinical intervention. Our clinically-deployable, markerless, accurate, effective, and efficient tool may be an effective weapon in combating the epidemic of non-contact ACL injury in female adolescent athletes.
References


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Appendix A

Processing Code

import SimpleOpenNI.*;
import ddf.minim.*;
SimpleOpenNI kinect;

//minim sound objects
Minim minim;
AudioSnippet player;

PrintWriter output;

String timestamp;
void setup() {
size(640, 480);
stroke(255, 0, 0);
strokeWeight(4);
textSize(20);

timestamp = year() + nf(month(),2) + nf(day(),2) + "," + nf(hour(),2) + nf(minute(),2) + nf(second(),2);

output = createWriter((timestamp)+"angles.txt");
frameRate(60);

kinect = new SimpleOpenNI(this);
kinect.enableDepth();
kinect.enableUser(SimpleOpenNI.SKELETON_PROFILE_ALL);
kinect.setMirror(true);

//initialize Minim and Audioplayer
minim = new Minim(this);
player = minim.loadSnippet("beep.wav");
}

void draw() {
kinect.update();
PImage depth = kinect.depthImage();
image(depth, 0, 0);
frame.setTitle(int(frameRate) + " fps");
}
IntVector userList = new IntVector();
kinect getUsers(userList);

if (userList.size() > 0) {
    int userId = userList.get(0);

    if (kinect.isTrackingSkeleton(userId)) {
        // get the positions of the three joint points for each knee
        PVector rightFoot = new PVector();
        kinect.getJointPositionSkeleton(userId,
            SimpleOpenNI.SKEL_RIGHT_FOOT, rightFoot);

        PVector rightKnee = new PVector();
        kinect.getJointPositionSkeleton(userId,
            SimpleOpenNI.SKEL_RIGHT_KNEE, rightKnee);

        PVector rightHip = new PVector();
        kinect.getJointPositionSkeleton(userId,
            SimpleOpenNI.SKEL_RIGHT_HIP, rightHip);

        PVector leftFoot = new PVector();
        kinect.getJointPositionSkeleton(userId,
            SimpleOpenNI.SKEL_LEFT_FOOT, leftFoot);

        PVector leftKnee = new PVector();
        kinect.getJointPositionSkeleton(userId,
            SimpleOpenNI.SKEL_LEFT_KNEE, leftKnee);
PVector leftHip = new PVector();
kinect.getJointPositionSkeleton(userId,
    SimpleOpenNI.SKEL_LEFT_HIP, leftHip);

// convert our knee joint points into screen space coordinates
PVector convertedRightFoot = new PVector();
kinect.convertRealWorldToProjective(rightFoot, convertedRightFoot);

PVector convertedRightKnee = new PVector();
kinect.convertRealWorldToProjective(rightKnee, convertedRightKnee);

PVector convertedRightHip = new PVector();
kinect.convertRealWorldToProjective(rightHip, convertedRightHip);

PVector convertedLeftFoot = new PVector();
kinect.convertRealWorldToProjective(leftFoot, convertedLeftFoot);

PVector convertedLeftKnee = new PVector();
kinect.convertRealWorldToProjective(leftKnee, convertedLeftKnee);

PVector convertedLeftHip = new PVector();
kinect.convertRealWorldToProjective(leftHip, convertedLeftHip);

// reduce our joint vectors to two dimensions for valgus
PVector rightFoot2D = new PVector(rightFoot.x, rightFoot.y);
PVector rightKnee2D = new PVector(rightKnee.x, rightKnee.y);
PVector rightHip2D = new PVector(rightHip.x, rightHip.y);

// reduce our joint vectors to two dimensions for flexion
PVector rightFoot2Dflex = new PVector(rightFoot.y, rightFoot.z);
PVector rightKnee2Dflex = new PVector(rightKnee.y, rightKnee.z);
PVector rightHip2Dflex = new PVector(rightHip.y, rightHip.z);

// calculate the axes against which we want to measure our valgus angle
PVector thighOrientation = PVector.sub(rightHip2D, rightKnee2D);

// calculate the axes against which we want to measure our flexion angle
PVector thighOrientationflex = PVector.sub(rightHip2Dflex, rightKnee2Dflex);

// calculate the frontal plane angle of the knee
float shankAngle =
    angleOf(rightFoot2D, rightKnee2D, thighOrientation);

// calculate the sagittal plane angle of the knee
float shankAngleflex =
    angleOf(rightFoot2Dflex, rightKnee2Dflex, thighOrientationflex);

// reduce our joint vectors to two dimensions for valgus
PVector leftFoot2D = new PVector(leftFoot.x, leftFoot.y);
PVector leftKnee2D = new PVector(leftKnee.x, leftKnee.y);
PVector leftHip2D = new PVector(leftHip.x, leftHip.y);
// reduce our joint vectors to two dimensions for flexion
PVector leftFoot2Dflex = new PVector(leftFoot.y, leftFoot.z);
PVector leftKnee2Dflex = new PVector(leftKnee.y, leftKnee.z);
PVector leftHip2Dflex = new PVector(leftHip.y, leftHip.z);

// calculate the axes against which we want to measure our valgus angle
PVector thighOrientationL = PVector.sub(leftHip2D, leftKnee2D);

// calculate the axes against which we want to measure our flexion angle
PVector thighOrientationflexL = PVector.sub(leftHip2Dflex, leftKnee2Dflex);

// calculate the frontal plane angle of the knee
float shankAngleL =
    angleOf(leftFoot2D, leftKnee2D, thighOrientationL);

// calculate the sagittal plane angle of the knee
float shankAngleflexL =
    angleOf(leftFoot2Dflex, leftKnee2Dflex, thighOrientationflexL);

// calculate difference between hips by subtracting one vector from another
PVector differenceVectorHip = PVector.sub(leftHip2D, rightHip2D);

// calculate the distance and direction of the hip difference vector
float magnitudeHip = differenceVectorHip.mag();
differenceVectorHip.normalize();

// calculate difference between knees by subtracting one vector from another
PVector differenceVector = PVector.sub(leftKnee, rightKnee);

// calculate the knee distance and direction of the difference vector
float magnitude = differenceVector.mag();
differenceVector.normalize();

// draw a line between the knees
if (magnitude < magnitudeHip*1.5) {
    pushMatrix();
    stroke(255,0,0);
    kinect.drawLimb(userId, SimpleOpenNI.SKEL_LEFT_KNEE, SimpleOpenNI.SKEL_RIGHT_KNEE);
    popMatrix();
}

else {
    pushMatrix();
    stroke(0,255,0);
    kinect.drawLimb(userId, SimpleOpenNI.SKEL_LEFT_KNEE, SimpleOpenNI.SKEL_RIGHT_KNEE);
    popMatrix();
}

// display
if (magnitude < magnitudeHip*1.5) {
    //play beep.wav feedback sound when values are outside of desired threshold
    player.play();
    //display
    pushMatrix();
    fill(255,0,0);
text("Knee Separation (mm): " + int(magnitude), 20, 80);
popMatrix();

// if sound has played all the way through, rewind and repeat
if (!player.isPlaying()) {
    player.rewind();
    // player.pause();
}
else {
    pushMatrix();
    fill(0, 255, 0);
    text("Knee Separation (mm): " + int(magnitude), 20, 80);
    popMatrix();
}

// show the left flexion angles on the screen
scale(1);

if (int(shankAngleflexL) < 30) {
    // play beep.wav feedback sound when values are outside of desired threshold
    // player.play();
    // display
    pushMatrix();
    fill(255, 0, 0);
    text("L Knee Flexion (deg): " + int(shankAngleflexL) + "\n", 20, 50);
    popMatrix();
}
//if sound has played all the way through, rewind and repeat

//if (!player.isPlaying()) {
// player.rewind();
//player.pause();
//}

} else {
pushMatrix();
fill(0,255,0);
text("L Knee Flexion (deg): " + int(shankAngleflexL) + ", 20, 50);
popMatrix(); }

//text("L Knee Valgus (deg): " + int(shankAngleL) + ", 20, 20);

// show the right flexion angles on the screen
scale(1);
if (int(shankAngleflex) < 30) {
//play beep.wav feedback sound when values are outside of desired threshold
//player.play();
//display
pushMatrix();
fill(255,0,0);
text("R Knee Flexion (deg): " + int(shankAngleflex) + ", 360, 50);
popMatrix();
//if sound has played all the way through, rewind and repeat
// if (!player.isPlaying()) {
// player.rewind();


// player.pause();

else {
    pushMatrix();
    fill(0,255,0);
    text("R Knee Flexion (deg): " + int(shankAngleflex) + "\n", 360, 50);
    popMatrix();
}

//text("R Knee Valgus (deg): " + int(shankAngle) + "\n", 360, 20);

//draw skeleton
stroke(150,150,250);
kinect.drawLimb(userId, SimpleOpenNI.SKEL_RIGHT_HIP, SimpleOpenNI.SKEL_RIGHT_KNEE);
kinect.drawLimb(userId, SimpleOpenNI.SKEL_RIGHT_KNEE, SimpleOpenNI.SKEL_RIGHT_FOOT);
kinect.drawLimb(userId, SimpleOpenNI.SKEL_RIGHT_HIP, SimpleOpenNI.SKEL_LEFT_HIP);
kinect.drawLimb(userId, SimpleOpenNI.SKEL_LEFT_HIP, SimpleOpenNI.SKEL_LEFT_KNEE);
kinect.drawLimb(userId, SimpleOpenNI.SKEL_LEFT_KNEE, SimpleOpenNI.SKEL_LEFT_FOOT);

// Write the coordinate to a file with a
// "t" (TAB character) between each entry
output.println(millis() + "t" + int(shankAngleflexL) + "t" + int(shankAngleflex) + "t" + int(magnitude) + "t" + int(magnitudeHip));
void stop() {
    player.close();
    minim.stop();
    super.stop();
}

void keyPressed() { // Press a key to save the data
    output.flush(); // Write the remaining data
    output.close(); // Finish the file
    exit(); // Stop the program
}

float angleOf(PVector one, PVector two, PVector axis) {
    PVector limb = PVector.sub(two, one);
    return degrees(PVector.angleBetween(limb, axis));
}

// user-tracking callbacks!
void onNewUser(int userId) {
    println("start pose detection");
    kinect.startPoseDetection("Psi", userId);
}

// user-tracking callbacks!
void onNewUser(int userId) {
    println("start pose detection");
    kinect.startPoseDetection("Psi", userId);
}
void onEndCalibration(int userId, boolean successful) {
if (successful) {
    println(" User calibrated !!!");
    kinect.startTrackingSkeleton(userId);
}
else {
    println(" Failed to calibrate user !!!");
    kinect.startPoseDetection("Psi", userId);
}
}

void onStartPose(String pose, int userId) {
    println("Started pose for user");
    kinect.stopPoseDetection(userId);
    kinect.requestCalibrationSkeleton(userId, true);
}
Appendix B

Injury History / Activity Questionnaire
Activity Level & Injury History Questionnaire

1. Do you currently participate in regular exercise / athletics (i.e. ≥ 3 days/week)?
   □ Yes □ No

2. How many days a week do you participate in exercise / practice?
   □ 1-2 □ 3-4 □ 5-7

3. How long have you participated in regular exercise / your sport?
   □ < 6 weeks □ 6-8 weeks □ 8-12 weeks
   □ 3-6 months □ 6-12 months □ >12 months
   □ 12-24 months □ > 24 months

4. Have you had any injuries to your feet, legs, knees, hips, back, or neck that have caused you to more than three consecutive days of practice during the past 3 months?
   □ Yes □ No

5. Have you EVER had any broken bones in, or surgery on, your feet, legs, knees, hips, back, or neck?
   □ Yes □ No

6. Have you had any neurological injuries within the past 3 months (concussion, etc)?
   □ Yes □ No
# Appendix C

## Additional Statistical Tables

### Table C.1: Subject Data

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<td>1576.4</td>
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<td>5.16</td>
<td>1.63</td>
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<td>24</td>
<td>Fem</td>
<td>18</td>
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<td>63.08</td>
<td>8</td>
<td>Biofeed</td>
<td>261</td>
<td>363</td>
<td>141</td>
<td>25</td>
<td>8</td>
<td>7</td>
<td>-0.14</td>
<td>Normal</td>
<td>1576.4</td>
<td>21</td>
<td>5.16</td>
<td>1.63</td>
<td>-0.56</td>
</tr>
</tbody>
</table>
Table C.2: System Pearson Correlation for Pre-Training Knee Flexion

<table>
<thead>
<tr>
<th></th>
<th>MocapFlexPre</th>
<th>KinectFlexPre</th>
</tr>
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<tbody>
<tr>
<td>MOCAP FlexPre</td>
<td>Pearson Correlation</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Sig. (1-tailed)</td>
<td>.000</td>
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<tr>
<td></td>
<td>Sum of Squares and Cross-products</td>
<td>845.875</td>
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<tr>
<td></td>
<td>Covariance</td>
<td>120.839</td>
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<tr>
<td></td>
<td>N</td>
<td>8</td>
</tr>
<tr>
<td>Kinect FlexPre</td>
<td>Pearson Correlation</td>
<td>.963**</td>
</tr>
<tr>
<td></td>
<td>Sig. (1-tailed)</td>
<td>.000</td>
</tr>
<tr>
<td></td>
<td>Sum of Squares and Cross-products</td>
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<tr>
<td></td>
<td>Covariance</td>
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</table>

**. Correlation is significant at the 0.01 level (1-tailed).

Table C.3: System Pearson Correlation for Post-Training Knee Flexion

<table>
<thead>
<tr>
<th></th>
<th>MocapFlexPost</th>
<th>KinectFlexPost</th>
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</thead>
<tbody>
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<td>MOCAP FlexPost</td>
<td>Pearson Correlation</td>
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</tr>
<tr>
<td></td>
<td>Sig. (1-tailed)</td>
<td>.001</td>
</tr>
<tr>
<td></td>
<td>Sum of Squares and Cross-products</td>
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<td></td>
<td>Covariance</td>
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<td>N</td>
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<tr>
<td>Kinect FlexPost</td>
<td>Pearson Correlation</td>
<td>.897**</td>
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<td>Sig. (1-tailed)</td>
<td>.001</td>
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<td></td>
<td>Sum of Squares and Cross-products</td>
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<td></td>
<td>Covariance</td>
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**. Correlation is significant at the 0.01 level (1-tailed).
Table C.4: System Pearson Correlation for Pre-Training Knee Separation Distance

<table>
<thead>
<tr>
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<th>MocapK2KPre</th>
<th>KinectK2KPre</th>
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</thead>
<tbody>
<tr>
<td>MOCAP K2KPre</td>
<td>Pearson Correlation</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Sig. (1-tailed)</td>
<td>.007</td>
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<td></td>
<td>Sum of Squares and Cross-products</td>
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<td>Covariance</td>
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<td>Kinect K2KPre</td>
<td>Pearson Correlation</td>
<td>.815**</td>
</tr>
<tr>
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<td>Sig. (1-tailed)</td>
<td>.007</td>
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<td>Sum of Squares and Cross-products</td>
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**. Correlation is significant at the 0.01 level (1-tailed).

Table C.5: System Pearson Correlation for Post-Training Knee Separation Distance

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<th>MocapK2KPost</th>
<th>KinectK2KPost</th>
</tr>
</thead>
<tbody>
<tr>
<td>MOCAP K2KPost</td>
<td>Pearson Correlation</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Sig. (1-tailed)</td>
<td>.000</td>
</tr>
<tr>
<td></td>
<td>Sum of Squares and Cross-products</td>
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<td>Kinect K2KPost</td>
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<td>.961**</td>
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<td></td>
<td>Sig. (1-tailed)</td>
<td>.000</td>
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<td></td>
<td>Sum of Squares and Cross-products</td>
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<td></td>
<td>Covariance</td>
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**. Correlation is significant at the 0.01 level (1-tailed).
### Table C.6: Mean and SD for Knee Flexion Angle Change Scores

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Mean</th>
<th>Std. Deviation</th>
<th>Std. Error Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>ChgScoreFlex</td>
<td>12</td>
<td>-1.4167</td>
<td>2.77843</td>
<td>.80206</td>
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<td>Biofeedback</td>
<td>12</td>
<td>6.9167</td>
<td>5.77547</td>
<td>1.66723</td>
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</table>

### Table C.7: Independent t Test for Knee Flexion Angle Change Scores Between Groups

<table>
<thead>
<tr>
<th>t-test for Equality of Means</th>
<th>df</th>
<th>Sig. (2-tailed)</th>
<th>Mean Difference</th>
<th>Std. Error Difference</th>
<th>95% Confidence Interval of the Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>t</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>ChgScoreFlex Equal variances assumed</td>
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<td>.001</td>
<td>-7.33333</td>
<td>1.85013</td>
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</tbody>
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### Table C.8: Means and SD for Change in Knee Separation Distance

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Mean</th>
<th>Std. Deviation</th>
<th>Std. Error Mean</th>
</tr>
</thead>
<tbody>
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<td>NormChgScoreDist</td>
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<td>Biofeedback</td>
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<td>32.2500</td>
<td>21.23944</td>
<td>6.13130</td>
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### Table C.9: Independent t Test for Change in Knee Separation Distance Between Groups

<table>
<thead>
<tr>
<th>t-test for Equality of Means</th>
<th>df</th>
<th>Sig. (2-tailed)</th>
<th>Mean Difference</th>
<th>Std. Error Difference</th>
<th>95% Confidence Interval of the Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>t</td>
<td></td>
<td></td>
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<tr>
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<td>.024</td>
<td>-20.33333</td>
<td>8.41318</td>
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</table>
Appendix D

IRB Documentation
PARENTAL PERMISSION FORM

THE EFFECTS OF AN OPENNI / KINECT-BASED BIOFEEDBACK INTERVENTION ON KINEMATICS AT THE KNEE DURING DROP VERTICAL JUMP LANDINGS: IMPLICATIONS FOR REDUCING NEUROMUSCULAR PREDISPOSITION TO NON-CONTACT ACL INJURY RISK IN THE YOUNG FEMALE ATHLETE

Principal Investigator: Dr. Barry W. Scheuermann, Ph.D.
Other Staff: Charles W. Armstrong, Ph.D., Coordinator
Edward Nyman, Co-Investigator

Contact Phone number(s): (419) 583-0189

PURPOSE (WHY THIS RESEARCH IS BEING DONE)
Your child is being asked to take part in a research study of jump landing mechanics. The purpose of the study is to determine the effectiveness of a new approach to accurately measure landing mechanics while providing visual feedback on proper landing mechanics in real-time for the athlete.

Your child was selected as someone who may want to take part in this study because of her age, gender, and activity level, as one of approximately forty (40) total subjects in the study.

DESCRIPTION OF THE RESEARCH PROCEDURES AND DURATION OF YOUR CHILD'S INVOLVEMENT
If you decide for your child to take part in this study, your child will be asked to jump off of a short box and land on the floor with both feet while a small camera records the movement of her knees. The first time she uses the system, we will give her an introduction to how it works and then measure the characteristics of her landing.

She, and other participants, will use this system as a training device for about 10 minutes over the course of which time she will complete multiple jumps from the box, three times per week during normal practice times. At the end of the training period, the characteristics of her landing will be measured several times and compared to the responses recorded on the first day.

*Permission to record: Will you permit the researcher to video record during this research procedure?

YES [ ] NO [ ]

Initial Here

UNIVERSITY OF TOLEDO IRB
APPROVAL DATE: 06/17/2013
EXPIRATION DATE: 06/16/2014
This space for IRB approved Data Stamp
RISKS AND DISCOMFORTS YOUR CHILD MAY EXPERIENCE IF YOUR CHILD TAKES PART IN THIS RESEARCH
The foreseeable risks, though not likely, to this study are the chance of musculoskeletal injury including the following:

- Muscle Soreness
- Ankle Sprain
- Knee Sprain
- Strain Muscles in Lower Extremity
- Falls

Other possible, but unlikely, risks include fracture to regions within the lower extremities (ankle, leg).

POSSIBLE BENEFIT TO YOUR CHILD BY TAKING PART IN THIS RESEARCH
Possible benefits include an enhanced knowledge of jump landing mechanics that may potentially lead to an improvement in the prevention of lower extremity injury risk. However, we cannot and do not guarantee or promise that your child will receive any benefits from this research.

COST TO YOU OR YOUR CHILD FOR TAKING PART IN THIS STUDY
There is no cost to you for your child's participation in this study.

PAYMENT OR OTHER COMPENSATION TO YOUR CHILD FOR TAKING PART IN THIS RESEARCH
If you decide for your child to take part in this research, your child will receive no monetary compensation.

ALTERNATIVE(S) TO TAKING PART IN THIS RESEARCH
The alternative is not to participate in this study and continue to receive the same training currently provided by your child's sports programming.

CONFIDENTIALITY
The researchers will make every effort to prevent anyone who is not on the research team from knowing that you provided this information, or what that information is. The consent forms with signatures will be kept separate from the information we collect, which will not include names and which will be presented to others only when combined with other responses. Although we will make every effort to protect your confidentiality, there is a low risk that this might be breached.

IN THE EVENT OF A RESEARCH-RELATED INJURY
In the event of injury resulting from your child's taking part in this study, treatment can be obtained at a health care facility of your choice. You should understand that the costs of such treatment will be your responsibility. Financial compensation is not available through The University of Toledo or The University of Toledo Medical Center.

By signing this form you are not giving up any of your legal rights as the parent of a research subject.

In the event of an injury, immediately contact Ed Nyman at 419-583-0189.

Continued on Next Page
VOLUNTARY PARTICIPATION
Taking part in this study is voluntary. You may refuse to allow your child to participate or discontinue participation at any time without penalty or a loss of benefits to which your child is otherwise entitled. If you decide for your child not to participate or to discontinue participation, your decision will not affect your future relations with the University of Toledo, The University of Toledo Medical Center, or your child's training facility.

NEW FINDINGS
You will be notified of new information that might change your decision for your child to be in this study if any becomes available.

Continued on Next Page
OFFER TO ANSWER QUESTIONS
Before you sign this form, please ask any questions on any aspect of this study that is unclear to you. You may take as much time as necessary to think it over. If you have questions regarding the research at any time before, during or after the study, you may contact Ed Nyman, 419-533-0180.

If you have questions beyond those answered by the research team or your child’s rights as a research subject or research-related injuries, please feel free to contact the Chairperson of the University of Toledo Biomedical Institutional Review Board at 419-383-6796.

SIGNATURE SECTION (Please read carefully)
YOU ARE MAKING A DECISION WHETHER OR NOT YOUR CHILD WILL PARTICIPATE IN THIS RESEARCH STUDY. YOUR SIGNATURE INDICATES THAT YOU HAVE READ THE INFORMATION PROVIDED ABOVE, YOU HAVE HAD ALL YOUR QUESTIONS ANSWERED, AND YOU HAVE DECIDED FOR YOUR CHILD TO TAKE PART IN THIS RESEARCH.

BY SIGNING THIS DOCUMENT YOU AUTHORIZE US TO USE OR DISCLOSE YOUR CHILD’S PROTECTED HEALTH INFORMATION AS DESCRIBED IN THIS FORM.

The date you sign this document to enroll in this study, that is, today’s date, MUST fall between the dates indicated on the approval stamp affixed to the bottom of each page. These dates indicate that this form is valid when you enroll in the study but do not reflect how long your child may participate in the study. Each page of this Consent/Authorization Form is stamped to indicate the form’s validity as approved by the UT Biomedical Institutional Review Board (IRB).

Name of Subject (please print) Signature of Subject or Person Authorized to Consent Date

Relationship to the Subject (Healthcare Power of Attorney authority or Legal Guardian) Time a.m. p.m.

Name of Person Providing Consent (please print) Signature of Person Providing Consent Date

YOU WILL BE GIVEN A SIGNED COPY OF THIS FORM TO KEEP.
CHILD RESEARCH SUBJECT ASSENT FORM

The Effects of an OpenNI / Kinect-Based Biofeedback Intervention on Kinematics at the Knee During Drop Vertical Jump Landings

Principal Investigator: Dr. Barry W. Schueremans, Chair, Department of Kinesiology 419-530-2692
Dr. Charles W. Armstrong, Coordinator
Edward Nyman, Co-Investigator 419-583-0189

- You are being asked to be in a study to help understand people better.
- You should ask any questions you have before making up your mind. You can think about it and discuss it with your family or friends before you decide.
- It is okay to say "No" if you don't want to be in the study. If you say "Yes" you can change your mind and then quit the study at any time without any problems.

We are doing a research study about the way young athletes land when jumping. A research study is a way to learn more about people. If you decide that you want to be part of this study, you will be asked to jump off a low box and land while a small camera records the way your knees move. You'll be able to see a live picture of the activity as it is happening.

Not everyone who takes part in this study will benefit. A benefit means that something good happens to you. We think these benefits might be learning more about how you jump and learning to land more safely.

When we are finished with this study we will write a report about what was learned. This report will not include your name or say that you were in the study.

If you have any questions about the study, you can ask Ed Nyman. You can call the investigators listed at the top of this page if you have a question later.

If you choose to be in this study, you agree to participate in the short testing and training jumps included, during your regularly scheduled practice times, a few days per week for six weeks.

CONTINUED NEXT PAGE
You do not have to be in this study if you do not want to. You can decide later if you want to think about it for awhile. If you decide to be in this study, please print and sign your name below.

I, ________________________________________, want to be in this research study.

(Print your name here)

Sign your Name: ____________________________ Date: ____________________________

Name of Person Explaining Assent (Print)

Signature of Person Explaining Assent Date

I attest that I or my representative discussed this study with the above-named participant.

Signature of Principal Investigator or Sub-Investigator Date