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The Influence Of Sex And Body Size On The Validity Of The Microsoft Kinect For Measuring Knee Motion During Landing

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Abstract

Measuring knee motion during landing is a method to evaluate knee injury risk. Three-dimensional motion capture is inaccessible, and the Microsoft Kinect is an alternative to measure knee motion. The primary objective was to evaluate the influence of sex and body size on the validity of the Kinect to measure knee motion during landing. A secondary objective was to compare knee motion between females and males with high and low body mass index (BMI). We assessed frontal plane knee kinematics of 40 (10 per group of females and males with high and low BMI) participants during landing with the Kinect and 3D motion capture. Good agreement between methods was found for the knee ankle separation ratio across groups, but there was low agreement between methods for measuring knee abduction. The high BMI group regardless of sex had more knee abduction than the low BMI group when measured with motion capture.

Keywords

ACL injury

Motion capture

Sex

Body Mass Index

Knee ankle separation ratio

Knee abduction angle

Knee valgus

Summary for Lay Audience

Traumatic knee injuries are common and result in missed playing time and long-term repercussions. The anterior cruciate ligament (ACL) is a knee structure that is important for stability during athletic tasks. ACL injuries require surgery and extensive physical rehabilitation. After rehabilitation, many athletes do not return to the same level of participation and are more likely to develop joint disease later in life.

Females have a higher incidence of ACL injury than males, which is partly due to physical characteristics such as bone shape, lower muscular strength, and different movement patterns. Additionally, females carry body mass around their hips and thighs compared with males who carry mass around the trunk. Therefore, females with greater mass may be disproportionately affected by poor movement patterns during landing. Individuals with larger body size have an increased risk of ACL injury. Furthermore, differences in body shape due to sex or size influence methods of assessing knee motion during landing.

This study evaluated the validity of a cost-effective and portable sensor that measures knee motion during landing relative to laboratory-based methods. There were two aims to the research: (1) to assess agreement between the portable sensor and laboratory-based motion capture for sex and body size groups, and (2) to compare knee motion during landing between females and males of different body sizes. We found that the sensor agreed with the motion capture for assessing knee position relative to the ankles, but not for measuring the angle between the thigh and shin. Furthermore, body size but not sex influenced the accuracy of the portable sensor. We also found that individuals with greater body size landed with more inward knee motion than individuals with lower body size.

The findings from this study may guide the use of the portable sensor for identifying individuals at greater risk of knee injury. Exercise programs have shown success in lowering knee injury rates by improving strength and coordination. Therefore, the portable sensor may be useful for identifying individuals who may benefit from

preventative exercises, but caution is recommended when measuring knee motion in those with larger body size.

Co-Authorship Statement

A version of this thesis is being prepared for submission in manuscript format. Jillian Neufeld is lead author and assisted with development of the research question and protocol design, and led data collection, processing and writing the report. Dr. Derek Pamukoff is a co-author who led the development of the research question and protocol and assisted with processing the data and critical revision. Vital Nwaokoro is a co-author who assisted with data collection.

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Chapter 1

1 Introduction

Traumatic knee injuries are a common cause of decreased participation in sport and physical activity[1,2]. The anterior cruciate ligament (ACL) is an anatomical structure that contributes to knee stability[3]. Anterior cruciate ligament injury management frequently includes reconstructive surgery and up to 12 months of rehabilitation[4]. Long-term consequences such as re-injury, increased risk of joint degeneration, and decreased participation in physical activity are associated with traumatic knee injuries[1,5–8].

Aberrant landing biomechanics may contribute to an increased risk of ACL injury[9,10]. For instance, dynamic knee valgus during landing is a common mechanism of ACL injury[9–11]. Dynamic knee valgus is a multi-joint and multi-planar movement involving hip internal rotation and adduction, knee abduction, tibial external rotation, and foot pronation and external rotation[11]. Movement patterns during landing have been modified through the implementation of neuromuscular training programs[12,13]. The knee abduction angle contributes to risk of ACL injury and is the primary surrogate measurement for dynamic knee valgus as it involves the distal femur nearing the midline, and the distal tibia moving away from the midline[8,9,14]. The Drop Vertical Jump (DVJ) is a common task for assessing biomechanics during landing[14]. Furthermore, frontal plane knee kinematics during a DVJ have been associated with ACL injury[9,14]. Therefore, evaluating frontal plane knee kinematics during a DVJ may identify individuals who would benefit from preventative neuromuscular exercises[15–18]. The gold-standard method for analyzing knee motion during a DVJ is 3-dimensional (3D) motion capture [19]. However, 3D motion capture is inaccessible for most clinicians due to cost and training requirements[20]. A 2-dimensional (2D) measurement method may provide an accessible alternative to 3D motion capture for clinical use.

The Microsoft Kinect was developed for the Xbox 360 game console[21] and has been used to measure knee movement during landings with good reliability and validity[18,22]. The

Kinect with ACL Gold software measures the knee ankle separation ratio (KASR) and knee valgus (knee abduction angle) when the feet first contact the floor and at the point of peak knee flexion. The KASR is the difference between the distance between the knees and the distance between the ankles and provides a surrogate measurement for knee abduction[16]. The knee abduction angle is the relative angle between the thigh and shank in the frontal plane. Medial knee positioning identified by the KASR, and knee abduction angle are associated with dynamic knee valgus and ACL injury[9,23]. Therefore, KASR and knee abduction angle may be useful for analyzing landing biomechanics.

Analyzing landing biomechanics is important for individuals who have a greater risk for ACL injury due to demographic factors like sex and body size. Females are 4-6 times more likely to sustain an ACL injury than males[24,25], which may be due to modifiable characteristics like movement patterns[26,27] and body composition[28,29]. For example, females land with less knee flexion and more knee abduction, which place stress on the ACL[14,30,]. Females also have anthropometric differences compared to males, with more mass stored around the hips and thighs[31,32]. The difference in body shape between males and females influences knee loading patterns[33,34]. Mass stored around the hips, which are lateral to the knee joint centers, contributes to valgus torque at the knees and may contribute to increased risk of ACL injury in females[34,35]. Secondly, greater body mass is associated with greater peak tibiofemoral compressive force going through the knee during weight bearing[36,37]. Greater tibiofemoral compression places stress on the ACL during landing[28]. Greater body size is also correlated with increased body surface area[38], which may influence motion capture analyses. The Microsoft Kinect uses a scan of body surface area to estimate limb positions and joint centers[21]. Differences in body surface area due to body size may influence accuracy in the Kinect's joint center identifications. Finally, there may be an interaction between sex and body size due to mass distribution, and females have a stronger correlation between body size and body fat percentage than males, and different soft tissue distribution[31,39].

Previous evaluations of the Microsoft Kinect did not evaluate the effect of sex and body size on validity and reliability. Sex, body size, and the possible interaction between sex and body size are important considerations for the validity of the Microsoft Kinect as they contribute to risk of ACL injury and may influence motion capture analyses[29,40,41]. Females in all body mass index (BMI) categories have more soft tissue around the hips compared to males of equivalent BMI classification[32] , which may influence joint center identification. Therefore, the differences in body surface area between sexes and those of different body size could influence the reliability and validity of the Microsoft Kinect [18].

The purpose of this study was to evaluate the influence of sex and body size on the validity of the Microsoft Kinect to measure frontal plane knee motion during landing. It was hypothesized that knee valgus characteristics during the DVJ assessed using the Microsoft Kinect would be reliable and valid against 3D motion capture for all groups.

A secondary purpose was to compare frontal plane knee motion during landing between males and females and those with high and low BMI using the Microsoft Kinect and 3D motion capture. It was hypothesized that females and those with high BMI would demonstrate greater knee valgus during landing than males and those with low BMI, respectively.

Chapter 2

2 Literature Review

2.1 Introduction

A reliable, valid, and cost-effective screening tool is needed to identify athletes at risk for traumatic knee injury. The Microsoft Kinect sensor with ACL Gold software is a cost effective and user-friendly method of analysing dynamic knee valgus during landing[18,42]. The Kinect is a motion analysis system that uses data gathered from the Kinect V2 sensor bar, originally designed for the Xbox 360[21]. The purpose of this review is to explore the available information about the Kinect technology, the variables contributing to dynamic knee valgus and risk of anterior cruciate ligament (ACL) injury, long term implications of ACL injury, and current methods and limitations for measuring movement patterns.

2.2 Knee injury epidemiology

Knee injuries in sport are a significant cause of lost playing time and decreased participation in sport[1,2,43]. The ACL is a knee structure that contributes to knee stability and is affected by traumatic knee injuries[44]. Anterior cruciate ligament injuries often require reconstructive surgery to regain stability and restore function[45]. Griffin et al. report an estimated 250,000 ACL injuries, and over 100,000 ACL reconstruction surgeries annually in the United States[46]. Surgery is followed by approximately 12 months of rehabilitation[45]. A cost effectiveness study by Stewart et al. reports an average cost for operative treatment of a first time ACL injury, without additional complications, at $18,174 US[45]. Stewart et al. also reported an average cost of physical therapy following ACL reconstruction as $6737.00 US[45]. Anterior cruciate ligament injuries occur primarily in young, active individuals and one report specifies that over 50% of ACL injuries occur in individuals aged 15-25 who participate in sport or physical activity[46]. Along with activity participation, there are other factors that contribute to risk of ACL injury such as sex and body size. A 2019 systematic review of ACL injury epidemiology in athletes ranging from amateur to professional found that 1 in 29 female athletes and 1 in 50 male athletes sustained ACL injuries over a 25-year study period[47]. Finally, post-traumatic osteoarthritis (PTOA) and decreased knee function are common outcomes of ACL injury regardless of surgical intervention[5–7,48,49].

2.3 Anterior cruciate ligament anatomy

The knee is comprised of the tibiofemoral joint and the patellofemoral joint[3]. The articulating bony surfaces of the tibiofemoral joint are the femoral condyles and the tibial plateau[3] . The ACL is a ligament of the tibiofemoral joint and connects the distal femur and the proximal tibia (Figure 1)[3,44]. The femoral attachment of the ACL is the intercondylar notch of the femur, on the medial aspect of the lateral femoral condyle[3,44]. The fibers of the ACL run distally and medially to insert on the anteromedial intercondylar eminence of the tibial plateau[3,11,44]. The ACL is composed of two bundles of fibers: the anteromedial bundle attaching more medially and anteriorly than the posterolateral bundle[11]. The posterolateral bundle attaches nearer to the middle of the tibial plateau than the anterolateral bundle[44]. Due to the locations of its attachments and the direction of its fibers, the ACL restricts anterior tibial translation and internal tibial rotation[3,28,44]. Different angles of knee flexion place tension on the different bundles of the ACL for knee stability throughout the full range of motion[44] .

The menisci are cartilage rings on the medial and lateral surfaces of the tibial plateau[3]. Meniscal injury is a common concomitant injury with ACL injury[50]. The menisci provide cushioning in the joint and their concave shape provides a better fit for the convex femoral condyles than the relatively flat tibial plateau[3]. The shape of the menisci also contributes to assisting the ACL in restricting translation of the femur on the tibia[3].

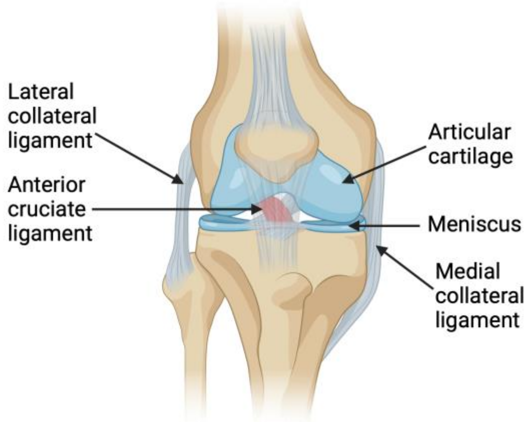

Figure 1: Structures of the knee. Created with BioRender.com

2.4 Knee valgus

An uncontrolled inward knee movement, into a "position of no return" during landings is a common mechanism of ACL injury[11]. The "position of no return" is a valgus position, characterized by an internally rotated and adducted hip, abducted knee, externally rotated tibia and pronated and externally rotated foot (Figure 2)[11].

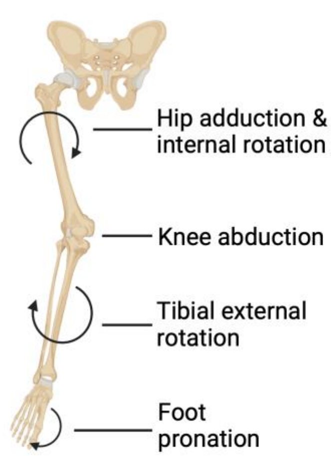

Figure 2: Components of knee valgus. Created with BioRender.com

One study suggests that the specific point during a landing when the valgus movement occurs is important as ACL tension is influenced by the knee flexion angle[51]. The knee flexion angle also influences the angle of pull of the quadriceps muscles, which contribute to anterior tibial translation and additional tension on the ACL[11,51]. Markolf et al. measured internal load on cadaver ACLs using a load transducer and showed that 15 Nm of external abduction torque applied to the knee contributed to an additional 200N pull from the quadriceps tendon and increased ACL stress by 41% at full extension, 100% at 10 degrees of knee flexion and 50% at 20 degrees of knee flexion[51]. The direct measure of ACL stress at different joint angles with additional external loads illustrates how medial knee positioning, flexion angle and quadriceps muscle pull interact to influence stress on the ACL[51].

Hewett, Meyer, and Ford used 3D motion capture to analyze landing mechanics during a drop vertical jump (DVJ) and compared between sexes and stages of adolescence[30]. They showed that females' total medial knee displacement increased between early and post pubertal stages and were greater than males' total medial knee displacement post puberty[30]. Moreover, post-pubescent females had a higher maximum knee abduction angle compared with males[30]. Another study by Hewett et al. completed a 3D analysis of a DVJ in 205 female athletes and tracked injury incidence over two soccer seasons and one basketball season[14]. Athletes who incurred ACL injuries during the study had greater knee abduction angles at initial contact and greater medial knee displacement compared to athletes who did not sustain ACL injuries[14]. Hewett et al. suggest that valgus knee movement during a dynamic task such as a DVJ is predictive of ACL injury[14,30]. Krosshaug et al. used 3D motion capture and force plates to measure knee abduction angles, knee flexion angles, vertical ground reaction forces, medial knee displacement and to calculate knee abduction moments through inverse dynamics during a DVJ[52]. The study tracked ACL injury incidence for 710 athlete participants from 2007-2014[52]. Medial knee displacement was the only variable that was associated with ACL injury, but the study reported low sensitivity and specificity, and recommended using medial knee displacement as an indicator of injury risk rather than a predictor of injury[52]. Paterno et al. conducted biomechanical landing analysis on athletes following ACL reconstruction surgery and followed them during their first year returning to sport[8]. Paterno et al. found increased hip internal rotation and greater medial knee displacement in the athletes who incurred a second ACL injury during the study[8]. Collectively, the body of research indicates that characteristics of dynamic knee valgus should be measured because they are important contributors to ACL injury and may vary between females and males.

2.5 Movement analysis application

Biomechanical analyses are used to screen athletes for injury risk, movement asymmetries and aberrant movement patterns[53–55]. Analysis tools such as the Kinect with ACL Gold, Functional Movement Screen (FMS) and Landing Error Scoring System (LESS) use visual analyses to assess movement patterns[42,53,56]. Biomechanical analyses have also been used to measure changes in variables associated with ACL injury, such as knee abduction angles and internal tibial rotation[57]. Pre- and post- intervention biomechanical analyses have shown the efficacy of exercise programs in altering movement technique[57,58]. Cochrane et al. used biomechanical analyses before and after a balance training and machine weight intervention and showed an improvement in change-of-direction performance[57]. Specifically, there was a decrease in peak knee abduction and decreased peak tibial internal rotation, which are associated with risk of ACL injury[57].

A cost-analysis study of widespread implementation of exercise intervention regardless of risk factors compared with athlete screening through movement analysis and targeted intervention found that current screening methods were cost-prohibitive and exercise interventions for all athletes regardless of risk levels were more effective[59]. Despite cost-effectiveness, widespread intervention approaches are limited by coaches' knowledge and implementation of the preventative program as well as athlete compliance and effort[27]. Therefore, a cost-effective screening tool that measures biomechanical factors related to ACL injury risk could identify athletes who would benefit from exercise interventions.

2.6 Microsoft Kinect

2.6.1 Sensor validity

The Microsoft Kinect sensor was originally designed for the Xbox 360 video game console[21,60]. The Kinect sensor uses a 2D camera, an infrared light projector, and an infrared depth sensor to map the surface area of objects in its field, which identifies joint centers and limb segments[21,60]. The depth sensor determines depth distance using time of flight technology where the length of time between a pulse of light being emitted from the infrared light projector and the reflected light returning to the sensor provides a depth distance estimate[61]. Software developed by Microsoft inputs the 2D pixel coordinates from the camera and the corresponding depth distance from the infrared sensor into a random-forest-based decision tree which calculates the probability of each pixel being "joint" or "non-joint"[62]. The areas of densest concentration of pixels identified as probably "joint" are identified as joint centers[62]. The decision tree was developed through exposing the software to pre-labeled motion capture data[62]. Following joint location identification, joints are labeled in a hierarchal fashion by identifying the spine base as a central point on the body surface area with other joint labels following based on their position relative to that central point. Thus, the joint centers immediately inferior and lateral to the spine base are the hip joints, the knees are distal to the ipsilateral hip, and the ankles distal to the ipsilateral knees[62]. Kinematics are measured as relative angles between limb segments whose end points are defined by joint center location estimates[21].

In a study of joint center identification accuracy, Xu et al. aligned the coordinate system of a Vicon 3D motion capture system with the Kinect's sensor field[21]. Xu et al. measured the difference between the locations of the joint centers identified by 3D motion capture and the Kinect skeletal model and found that the difference varied according to the joint and the body position[21]. When comparing knee joint center locations from a standing posture, Xu et al. found an average difference of 63mm (SD 51) for the left knee and 67mm (SD55) for the right knee[21]. From a squat position, Xu et al. reported an average difference of 108mm (SD 40) in left knee joint locations and 118mm (SD 50) in right knee joint locations[21]. Xu et al. reported mean joint location discrepancies as 115mm (SD49) and 111mm (SD44) for left and right hips respectively. Xu et al. did not propose benchmarks for acceptable levels of difference in joint center locations, but recommended considering each joint's discrepancy individually, understanding that error in joint center identification may carry forward into calculations of kinematics[21]. Xu et al. also reported an average BMI of 23.8 (SD 2.0) for their participants and stated that their results were not transferrable to subjects with higher BMI[21].

2.6.2 Measuring medial knee movement

Researchers from the University of Missouri designed the "ACL Gold" application which uses data from the Kinect sensor to measure knee motion during landing[63]. The Kinect with ACL Gold identifies "knee valgus angles" in the frontal plane and Knee Ankle Separation Ratios (KASR) (Figure 3) at initial foot contact and peak knee flexion during landing[18,42]. Knee ankle separation ratios describe the ratio of the distance between the knees compared to the distance between the ankles[18]. A KASR of 1 indicates the knees are directly superior to the ankles, a score less than 1 indicates less distance between the knees than the ankles[16]. Thus, lower KASR scores indicate more medial knee positioning[16].

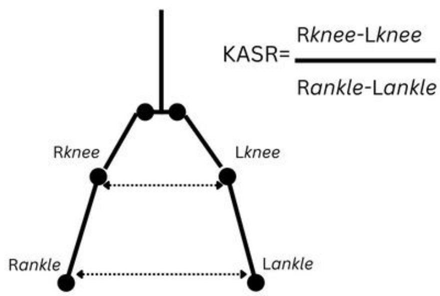

Figure 3: Knee ankle separation ratio (KASR)

Studies by Hewett et al. and Krosshaug et al. have shown the correlation between medial knee displacement and increased risk of ACL injury[9,14]. Hewett et al. used logistic regression analysis to examine the DVJ technique of athletes with ACL injuries to those without ACL injuries and found significantly higher peak knee abduction angles in the athletes with ACL injuries[14]. Limitations to the study by Hewett et al. include a limited sample of exclusively adolescent female soccer and basketball athletes. Hewett et al.'s analysis also considered each knee rather than participant as independent data points, potentially skewing the relationship between participants' movement patterns and knee injury. Despite the limitations of the study, the research illustrates the potential for implementing movement screening for injury incidence reduction. Koga et al. completed a case series examining knee positions in college-aged female athletes immediately prior to ACL injury during handball and basketball games[9]. The case series analyzed frames from footage of the injuries and used skeletal modelling to measure knee kinematics during the injury event[9]. Koga et al. found that during the 10 video frames prior to injury, all 10 of the case study's athletes experienced abrupt increases in knee abduction angles by an average of 12 degrees[9]. Nagano et al. used the Frontal Plane Projection Angle (FPPA) to measure knee valgus angles[64]. The frontal plane projection angle measures the angle between a line running through the Anterior Superior Iliac Spine (ASIS) and mid-patella and a line through mid-patella and the ankle joint[16,64]. Nagano et al. found agreement between 2D measures of the FPPA and 3D analysis of knee abduction angles for identifying athletes with higher risk of ACL injury based on a 9-degree knee abduction angle cut off proposed by Hewett et al.[14,64]. Following Nagano et al.'s research, Mizner et al. showed agreement between FPPA and KASR values for measuring knee valgus[16]. As such, the KASR may be a useful surrogate to measure ACL injury risk.

Gray et al. compared KASR measurements during a DVJ between the Kinect and Microsoft Gold and a Vicon 3D motion capture system[18]. Over 190 DVJ trials, Gray et al. showed that the Kinect system accurately identified initial contact and peak knee flexion 95.8% of the time[18]. However, accuracy was verified by visual evaluation of video frames and not inspected against force plate or 3D motion capture data[18]. The comparison of KASR measurements between the Kinect and 3D motion capture had an inter-class correlation (ICC) coefficient of 0.84 ($p<0.05$) at initial contact and 0.95 ($p<.05$) at peak knee flexion[18]. Gray et al. reported that they gathered sex and BMI information from their participants, but analyses were not stratified or reported by sex or BMI[18]. Gray et al. did not report agreement between systems for knee abduction angle[18].

In a similar study, Stone et al. evaluated agreement between the Kinect and 3D motion capture during a DVJ. To compare between methods, Stone et al. used the Microsoft Software development package to create their own measurement tool and aligned it with the 3D motion capture system[22]. Thus, measurements for both systems were extracted from frames identified by 3D motion capture data[22]. Stone et al. reported good-to-excellent agreement between the Kinect and 3D motion capture with ICC values of approximately 0.89 for both KASR and knee abduction angles[22]. In contrast, Tipton et al. compared measurements for peak knee valgus and peak knee flexion between the Kinect and a 3D motion capture system during a drop landing followed by single leg pivot task and found that the Kinect underestimated knee kinematics compared to the 3D analysis[65]. Tipton et al.'s analysis differed from previous comparisons between Kinect measurements and 3D motion capture because they analyzed a pivot, which changes the anatomical plane of the subject in the sensor's field[18,65]. Further, Tipton et al. cited a small sample size that was mostly male as limitations of their study[65]. Since the Kinect sensor uses the reflection off the surface of limbs to determine joint centers, different anthropometrics attributable to BMI and anatomical sex are a potential source of error in measuring medial knee movement.

2.7 Factors contributing to risk of knee injury

2.7.1 Sex related factors

Females have a greater likelihood of ACL injury than males per season of high school sport[66]. A systematic review of ACL injury incidence by Gornitzky et al. found that female athletes had a 0.66% chance of ACL injury per season compared to 0.42% chance of ACL injury per season for male athletes, which equates to a 48% greater likelihood for females compared to males[66]. The review also found that women's soccer had the highest risk of ACL injury per athletic exposure[66]. In a sport and athletic exposure matched study, females were 1.6 times more likely to sustain an ACL injury than males[25]. A longitudinal study of United States Marines showed that female Marines had a non-contact ACL injury incidence rate of 6.6% compared to 2.1% in male Marines, or nearly a 3:1 female-to-male ACL injury incidence[67].

Sex-specific risk factors that may influence the female-to-male ACL injury incidence discrepancy include non-modifiable structural differences such as quadriceps angle, ligament size and intercondylar notch width[28,67–69]. Modifiable sex-related variables include strength, neuromuscular control, and body mass index (BMI)[26,27,39,70].

2.7.2 Sex related structural differences

The quadriceps angle (Q-angle) is the angle between the line from the ASIS to the center of the patella and the line from the center of the patella to the tibial tuberosity[68]. A normal Q-angle for females is 2.7 to 5.8 degrees higher than males[71]. Higher Q-angles in females are attributed to larger pelvic width, greater tibiofemoral angles and increased structural femoral anteversion in females compared to males[68]. A study of skeletal alignment and knee joint loading by Hsu et al. found that females had greater anatomical Q-angle and tibial plateau valgus tilt than males[34]. Hsu et al. also found that females had higher maximum pressure on the tibial plateau than males when normalized for body weight[34]. Hsu et al used computer modeling to determine the effect of altered weight distribution and found that body mass located lateral to the knee joint center, as found in females with wider hips, contributed to a larger knee abduction moment[34]. Therefore, accurate characterizations of sex-specific anatomy are essential when utilizing motion capture analyses.

2.7.3 Strength differences between sexes

Muscular strength contributes to knee stability[72]. The main muscle groups acting on the knee joint are the hamstring muscles posteriorly and the quadriceps muscles anteriorly. The hamstring muscle group provides a posterior pull on the tibia, which prevents anterior tibial translation and reduces ACL strain[72]. The quadriceps' main action is knee extension, which pulls the tibia anteriorly and increases stress on the ACL[3,72,73]. Weak hamstring relative to quadriceps strength contributes to greater anterior translation of the shank, and greater ACL strain[70,72]. Strength ratios between hamstring and quadriceps force production are lower in females compared with males[70]. Both males and females experience an increase in muscular strength during puberty[70]. However, assessments of the change in quadriceps and hamstring strength have shown a disparity between the anterior and posterior muscle groups in females more than males[70]. Ahmad et al. showed that females' quadriceps strength increased to a greater extent than their hamstring strength during puberty, resulting in more anterior pull on the tibia from the quadriceps compared to posterior force from the hamstrings[70]. Conversely, males did not show the same disproportionate strength development in the anterior and posterior muscle groups[70].

The gluteal musculature also contributes to knee stability[74]. The gluteal muscles attach on the femur and the iliotibial band, which crosses the lateral knee to insert on the lateral condyle of the tibia. Strength and control of the gluteal muscles stabilize the femur and knee, and limit valgus movement and femoral internal rotation[74]. Suzuki et al. reported lower hip abduction and external rotation strength in females compared to males relative to body weight[75]. The study by Suzuki et al. also reported significant between sex differences in single-leg landing mechanics and showed a negative correlation between hip strength and femoral internal rotation while landing for females[75]. Less hip strength was associated with greater femoral internal rotation during landing, which is a factor of dynamic knee valgus and contributes to risk of ACL injury[11,75].

2.7.4 Neuromuscular control

Neuromuscular control refers to the unconscious control of the magnitude and timing of muscle contraction to produce coordinated movement and maintain joint stability during movement[72]. Coordinated contraction of the hamstrings, quadriceps and gluteal muscles increases joint compression and contributes to knee stability during landing by controlling the magnitude and velocity of movement[72]. Greater knee valgus positions have been associated with increased ACL injury risk[8,23]. Therefore, poor neuromuscular control of valgus knee motion contributes to increased risk of injury[8,76]. Females tend to have greater knee valgus during landing, which suggests a lower level of neuromuscular control compared to males[77]. An analysis of landings found a difference in the timing of maximal hip adduction and knee valgus during landing in females compared to males[78]. The analysis showed that females reached maximal hip adduction and maximal knee valgus before peak knee flexion, compared to males who reached maximal hip adduction and knee valgus after peak knee flexion[78]. The difference in knee positions during landing between females and males influences the amounts of stress on the ACL[51].

Ford et al. found that female athletes are more likely than males to rely heavily on their ligaments for stability rather than using their musculature to control and stabilize the joints during movement[77]. Ford et al. connected ligament dominance to a deficit in neuromuscular control in female athletes compared to their male counterparts[77].

Neuromuscular control can be modified through training[12,13,79]. A review of interventions aimed at ACL injury prevention conducted by Alentorn-Geli et al. reported a reduction in the difference in force absorption and joint stabilization between male and female athletes after the females participated in preventative training exercises[27]. Training neuromuscular control through exercise intervention programs has reduced ACL injury incidence[12,13,27,80]. Exercise interventions have also altered biomechanical variables related to ACL injury such as knee valgus movement during landing when compared to baseline measurements[12,13,27].

2.8 Body Mass Index

2.8.1 Body mass index classifications

Body mass index (BMI) uses height and body mass to classify body size[81]. The BMI scale ranges from underweight to extremely obese[81]. Overweight is defined as a BMI between 25.0-29.9kg/m^2 and obese is defined as a BMI >30.0 kg/m^2 [81]. Body mass index is related to body fat percentage and body surface area[39,82]. Body size influences movement analysis tools such as 3D motion capture and the Kinect with ACL Gold[40]. High BMI is also associated with poorer outcomes following knee injury[6,83], thus it is important to stratify movement analysis results by BMI.

2.8.2 Body mass index and primary injury

Higher body mass for a given height equates a higher BMI[81]. Higher mass results in increased axial load on weight bearing joints[36]. An axial load going through the knee joint increases anterior shear forces and internal tibial torques due to the shapes of the knee joint surfaces[28]. Increased anterior shear forces and internal tibial torques increase stress on the ACL[28].

Increased BMI is associated with an increased likelihood of concomitant knee injuries like meniscal tears and cartilage lesions in the knee, which may require additional surgical interventions at the time of reconstruction[83]. A review of the National College of Surgeon's National Surgical Quality Improvement database found that there was a positive association between increasing BMI and additional surgical procedures required at the time of ACL reconstruction[83]. Meniscal tears and cartilage lesions also contribute to the development of PTOA [5–7,49,83].

2.8.3 Body mass index and long-term joint health

Poorer outcomes and long-term repercussions of ACL injury in individuals with obesity indicates a need for a valid tool for screening individuals with a high BMI for aberrant movement patterns and increased risk of ACL injury[6,83]. Post-traumatic knee osteoarthritis is a secondary form of osteoarthritis, which occurs following a primary pathology such as ACL injury or meniscal injury[6,84,85]. A meta-analysis of studies tracking PTOA incidence reported PTOA development in 20.6% of ACL reconstructed patients within 10 years of reconstruction and 51.6% of patients within 20 years of reconstruction[84]. A study of PTOA 10 years after ACL reconstruction found PTOA in 87% who underwent ACL reconstruction with meniscal excision[85]. In comparison, non-specific osteoarthritis incidence is estimated at 19% in all adults over 45 years old[86].

Characteristics of PTOA include damage and loss of the articular cartilage, formation of osteophytes, and joint-space narrowing[85]. A review of PTOA cases within 5 years of ACL reconstruction found risk factors for PTOA development within that time frame included female sex (odds ration 1.2, p=0.002), obesity (odds ratio 1.,4 p=0.0004), and morbid obesity (odds ratio 1.5, p=0.0006) as compared to overall OA incidence[6].

Multiple mechanisms associated with ACL injury and BMI contribute to the development of PTOA such as increased joint load, decreased joint stability, and increased inflammatory characteristics[33,84,87,88]. Obesity is associated with higher compressive forces and contributes to deterioration of the knee cartilage and decreased joint space[33]. Decreased joint stability following ACL injury alters joint loading patterns in the knee and contributes to joint surface degeneration[88]. Anterior cruciate ligament reconstruction reduces joint laxity, but it does not restore pre-injury joint kinematics during weight bearing activity[88]. Thus, joint degeneration is seen in both reconstructed and ACL deficient knees, although to a lesser extent in reconstructed knees[89].

Obesity has been linked with increased concentrations of inflammatory substances such as free fatty acids, reactive oxygen species cytokines and adipokines which have a catabolic effect on tissues in the joint[87]. The catabolic effect of inflammatory substances contributes to the development or acceleration of PTOA[87]. PTOA has long term effects including increased pain, decreased function, and is associated with increased risk of cardiovascular disease and diabetes[90,91]. Individuals with obesity experience worse outcomes after ACL reconstruction according to the Knee Injury and Osteoarthritis Outcome Score (KOOS) and the Western Ontario and McMaster University Arthritis Index (WOMAC)[83].

2.8.4 Body mass index and body surface area

The Kinect sensor uses a scan of body surface area to identify joint locations for analyzing movement[21]. However, the distribution of body surface area changes as BMI increases [82]. A study of anthropometric measures including BMI and waist circumference compared these measurements between sexes to body fat measurements using the gold standard Dual-Energy X-ray Absorptiometry[39]. BMI classification was correlated with body fat percentage for both males and females for all age groups[39]. Furthermore, the correlation between BMI classification and percent body fat was stronger for females than males for every age group other than >80 years old, where it was the same between sexes[39]. The stronger correlation between BMI and body fat percentage for females indicates that higher BMI is related to higher levels of body fat in females[39].

A study of MRI images of the distal femur showed that subcutaneous fat thickness around the knee is correlated with BMI[38]. The study divided the cross-sectional image of the distal femur into four quadrants: posterolateral, posteromedial, anterolateral, and anteromedial[38]. The subcutaneous fat thicknesses in all four quadrants were predictors of BMI classification, with the posteromedial quadrant having the strongest correlation[38]. A study of subcutaneous fat thickness around the hip, measured through computed tomography, found that females had higher lateral fat thicknesses than males for all BMI groups and higher fat thickness ratios (lateral to anterior) than males across all BMI classifications[32]. Based on their measurements, Sprowls et al. attributed greater fat thickness ratios in females than males to higher lateral thickness levels in females rather than lower anterior thicknesses in males[32]. Increased mass lateral to the knee joint center increases the abduction moment at the knee[34]. Additionally, the location and thickness of body fat influences body surface area, which may confound estimates of joint center locations [32,38]. The accuracy of the Kinect's joint center estimates may vary between sexes and BMI classifications due to the associated differences in body surface area.

2.9 Current screening tools

2.9.1 Functional movement screen

The Functional Movement Screen (FMS) is a visual screening tool where observers score subjects' technique as they complete specific movements[92]. The FMS consists of 7 movements that are each scored from 0 to 3 (0 = painful, 1 = unable to complete, 2 = compensation required to complete movement, 3 = able to perform movement)[92] . A score equal or less than 14 is correlated with a history of injury[92,93]. The 7 movements include a deep squat, hurdle step, in-line lunge, shoulder mobility, active straight leg raise, trunk stability push-up and rotary stability[92]. The 7 movements differ from knee injury screening tests that use a DVJ as there is no landing component to any of the FMS movements.

A systematic review of studies assessing FMS reliability and validity found that FMS scoring is consistent between raters and within raters in repeated measures tests[92]. However, Kazman et al. questioned the predictive usefulness of the FMS score because the composite from the 7 movements is ambiguous[94]. Since the FMS test does not test for a particular injury or faulty movement pattern, the justification for combining the factors into a composite score to predict injury is weak[94]. Based on Kazman et al.'s research, clinicians should evaluate the movement patterns of each FMS task independently to assess athletes' strengths and weaknesses[94]. Conversely, Pamukoff et al. found that composite FMS scores were related to kinematic measurements during a DVJ in adults with a history of ACL reconstruction including the knee abduction angle[95]. Pamukoff et al. suggest further research on whether a modified FMS assessment of the overhead squat, inline lunge and hurdle step may be more related to the DVJ and may have a stronger association with injury[95]. The study also recommends using the FMS as one element of a battery of tests including a DVJ and further biomechanical analysis[95]. Unlike a composite FMS score, a 3D motion capture assessment of a DVJ quantifies knee kinematics and kinetics associated with risk of ACL injury[15].

2.9.2 Landing Error Scoring System

The Landing Error Scoring System (LESS) is an assessment for evaluating landing technique where an assessor grades the landing of a DVJ according to a list of errors [53,56]. The LESS is a visual evaluation of a DVJ from video recordings from the sagittal and frontal planes[96]. For each of the 17 errors on the scale, the assessor marks if the error was “present” for a score of 1, or “absent” for a score of 0, thus a higher score indicates more errors[53].

A study assessing the efficacy of the LESS for identifying athletes with a higher risk of ACL injury assessed 829 adolescent soccer players and followed up over the next 3 soccer seasons to track injuries[56]. In the study, Padua et al. showed that a LESS score of 5 was the defining line between lower and higher risk athletes[56]. Padua et al. found that the LESS was accurate in that the 7 athletes who sustained ACL injuries during the study did have LESS scores above the cut-off score of 5[56]. The low incidence of ACL injuries in the study (7 injuries over 3 years) made the positive predictive value of the LESS assessment 1.4%[56]. However, the association between scores greater than 5 and ACL injury indicates the potential usefulness of biomechanical screening for ACL injury risk[56].

A similar study by Smith et al. measured the relationship between LESS scores and subsequent ACL injury in high school and college athletes across multiple sports. Smith et al. found that intra-rater and inter-rater reliability of the LESS was high[53]. However, there was not a relationship between LESS scores and subsequent ACL injury[53].

A comparison of visual LESS scores and 3D motion capture analysis found that the strength of agreement between the visual scores and the 3D analysis differed between factors[97]. Onate et al. found high agreement between the visual LESS scores for knee valgus and knee flexion compared to the 3D measurements, but low agreement for sagittal plane knee angle at initial contact between the two methods[97]. Onate et al. suggested that the LESS requires further editing to eliminate factors with poor agreement with gold standard 3D motion capture while maintaining the factors most relevant to injury risk[97]. Importantly, the LESS does not quantify knee abduction angle, and thus, cannot replace motion capture methods that measure joint kinematics.

2.9.3 Gold standard 3-dimensional motion capture

2.9.3.1 Laboratory-based motion capture accessibility

Three-dimensional motion capture technology has been established as the gold standard in research for analyzing movement and biomechanics. 3D motion capture systems consist of a set of cameras that are calibrated to identify the position of reflective markers within the cameras' field of view. The positional information gathered by the cameras is translated into a digital representation of the markers, which are then labelled and combined with participant anthropometric data to create a skeletal model. Joint kinematics are derived from tracking changes in the locations of the joint centers and limb positions of the skeletal model.

Despite being the gold standard of motion analysis, 3D motion analysis is not widely utilized outside of laboratories and research facilities. The cost and training necessary to use the equipment are barriers to the widespread use of 3D motion capture[18]. Motion capture systems range in price from $100,000-$150,000 or more[18]. Along with technology expenses, data collection and reduction require training and time[23]. An accessible and cost-effective tool for assessing knee biomechanics could identify high risk individuals and be used for tracking changes in landing patterns over the course of strengthening or rehabilitation programs. The Kinect may provide objective measurements to guide individualization of interventions or guide return to play readiness.

2.9.3.2 3D motion capture marker sets

A common feature of 3D motion capture is the use of reflective markers, which are placed on specific anatomical landmarks to identify joint centers and construct limb segments[19,40,98,99]. The accuracy of limb segment and joint center identification is dependent on the marker placement. Therefore, inaccuracy or inconsistency in marker placement are sources of error in 3D motion analysis[19,40,98–100].

The Conventional Gait Model (CGM), also referred to as the Helen Hayes model, is an established method for defining limbs and segments of the human body in relation to one another[99–102]. The CGM estimates hip joint centers based on pelvic width, determined by the measurement between the ASIS markers and leg length[102]. Knee joint centers are defined as the perpendicular intersection of a line descending from the hip joint center and the midway point between the lateral and medial femoral condyle markers[102]. Ankle joint centers are derived from the assumption that the knee joint center, lateral malleolar marker and the halfway point between lateral and medial malleolar markers form a right triangle in the frontal plane[102]. The ratios the CGM uses to identify joint centers are applied universally and do not consider sex specific anatomical differences. The CGM is considered a hierarchical model because each joint center relies on the joint center of the segment immediately proximal as a reference point[102]. Therefore, error in marker placement will impact the calculations of adjacent joint center locations[102].

An alternative marker set to the CGM is the Six Degrees of Freedom (6DOF) set, which defines each body segment individually from the adjacent segments, allowing for individual segment analysis[101,103]. A study comparing the CGM marker set with the 6DOF marker set found better repeatability in marker placement for the CGM marker set as the higher 6DOF had a higher Standard Deviation (SD) error[101]. Despite being less repeatable than the CGM set, the 6DOF set met all clinical standards and allowed analysis of individual segments and planes of motion[101].

Regardless of the chosen marker set, using surface markers to define rigid limb segments and joint centers results in placement error from soft tissue[19,101,102]. When placing the markers, researchers and clinicians palpate anatomical landmarks to determine marker location[40,100–102,104]. Soft tissue can obscure bony landmarks resulting in misplaced surface markers[40,102]. Subcutaneous fat surrounding the knee is correlated with BMI classification, thus soft tissue could result in increased error in high BMI groups compared to low BMI groups[38,40]. In a hierarchical model like the CGM, increased subcutaneous fat surrounding the knee joint will influence knee joint center [102]. Error in limb segment and joint center identification from soft tissue will result in error in calculations of joint angles and kinematics[40,102]. Soft tissue may also affect joint center identification differently for females than males. Different body mass storage in the hips and pelvis in females may contribute to greater marker placement error from soft tissue in females than males.

Another error resulting from soft tissue is skin movement artifact[104]. Skin movement artifact is the difference between the measured movement of the surface marker and the true rigid segment movement[104]. As the limb segment moves, the surface markers move along with the skin and soft tissue rather than the rigid structure of the underlying bone. The difference between the marker movement due to skin movement artifact and the true rigid limb segment may cause over or underestimation of limb movement and result in miscalculated joint kinematics[104]. Critiques of the CGM include unfixed segment lengths, meaning that the limb segment length measured during the static trial is not held constant throughout movement trials[102]. Surface marker movement from skin movement artifact can alter limb segment length measurements over the course of a movement[102].

A study that examined the limitations of surface markers in 3D motion analysis used magnetic resonance imaging (MRI) to visualize femur location in comparison to the surface markers and found that the surface markers did not accurately represent the associated rigid limb segment[104]. The study found that the error was consistent for individuals, and within subject comparisons were still strong[104]. A comparison of surface marker and bone-pin marker use in 3D motion capture found that the magnitude of error between knee joint kinematics calculated using the bone-pin markers and using the surface markers was not consistent between subjects[104]. This between-subject difference suggests that individual body anthropometrics play a role in the magnitude of error due to skin movement artifact. The study suggests reporting all measurements derived from surface marker analysis with a standard error[104].

Lerner & Board investigated an obesity-specific surface marker set[40]. The goal of the research was to compare the established Helen Hayes surface marker set to an obesity-specific marker set to determine the effect of marker set choice on skin movement artifact[40]. The study grouped subjects by BMI and compared measurements for both the standard Helen Hayes marker set and an obesity-specific marker set for each group[40]. In the individuals without obesity, there was a high level of agreement between the Helen Hayes and obesity-specific marker sets[40]. However, in the group of individuals with obesity, there were significant measurement differences for hip flexion and pelvic tilt angles between the marker sets[40]. The different levels of agreement between marker sets in the groups with and without obesity demonstrate the influence of BMI on 3D motion analysis[40].

2.10 Summary

Anterior cruciate ligament injuries are traumatic injuries with high prevalence in young, active populations, affect females more than males and have significant long-term consequences, particularly for individuals in above normal BMI classes[5,45–47]. Valgus knee motion during landing is a common mechanism of ACL injury[77]. Screening for valgus knee movement, or factors associated with dynamic knee valgus during landing is a potential method for identifying individuals at increased risk of ACL injury[105]. Current gold standard 3D motion analysis technology is inaccessible to many practitioners due to high costs and training requirements[18]. The Kinect with ACL Gold software presents an accessible alternative to 3D motion capture, but validity and reliability may be influenced by sex and BMI.

Chapter 3

3 Methodology

This study was a single session, between methods study involving anthropometric measurements, a warm-up protocol, and a Drop Vertical Jump (DVJ) protocol.

3.1 Participants

Forty participants (10 low BMI female, 10 high BMI female, 10 low BMI male, 10 high BMI male) were recruited from the general student population. Sample size was based on an expected intraclass correlation (ICC) of 0.84, 5 measurements per participant, and a minimum acceptable ICC of 0.5 (Beta=0.2, alpha=0.05). Expected ICC was based on previous research evaluating the validity of the Kinect relative to 3D motion capture[18]. Participants were eligible if they were 18 years of age or older, able to read and comprehend English, and fall into a BMI category between 18.5 kg/m^2 – 24.9 kg/m^2 or >30.kg/m^2. Exclusion criteria included lower body injury within 6 months of participation, history of lower body surgery, or ongoing knee pain. Participants provided written consent and all methods were approved by the University Research Ethics Review Board.

3.2 Anthropometrics

Participants' height (meters) and mass (kilograms) were measured using a digital scale and wall mounted stadiometer respectively to determine body mass index [$BMI = \frac{mass\ (kg)}{height^2(m)}$]. Subjects were allocated to either the low BMI group [18.5-24.9kg/m^2], or high BMI [>30kg/m^2]. Individuals whose BMI fell between 25.0-29.9 kg/m^2 were excluded to ensure a considerable difference in body size between groups.

3.3 Instrumentation

An 8-camera motion capture system recording at 240Hz (Qualisys, Gothenburg Sweden) and force plates recording at 2400Hz (AMTI, Newton MA) were used for the 3D motion capture measurements. Following a general warm-up of body weight squats, lunges, and jumping jacks, participants were outfitted with reflective markers (Figure 4). Calibration markers were placed bilaterally on the iliac crests, greater trochanters, medial and lateral epicondyles of the knees, medial and lateral malleoli, the base of the first and fifth metatarsals, and the calcanei. Rigid marker clusters of 4 non-colinear reflective markers were affixed to the sacrum and bilaterally on the thighs, shanks, and dorsum of the feet for dynamic tracking. The use of rigid clusters has been shown to increase the reliability of the 3D motion capture in high BMI groups[40]. The Microsoft Kinect was positioned on a tripod 2.5m[18] in front of the force plate setup within the 8-camera field, perpendicular to the anterior-posterior axis to allow measurement within the frontal plane (Figure 5).

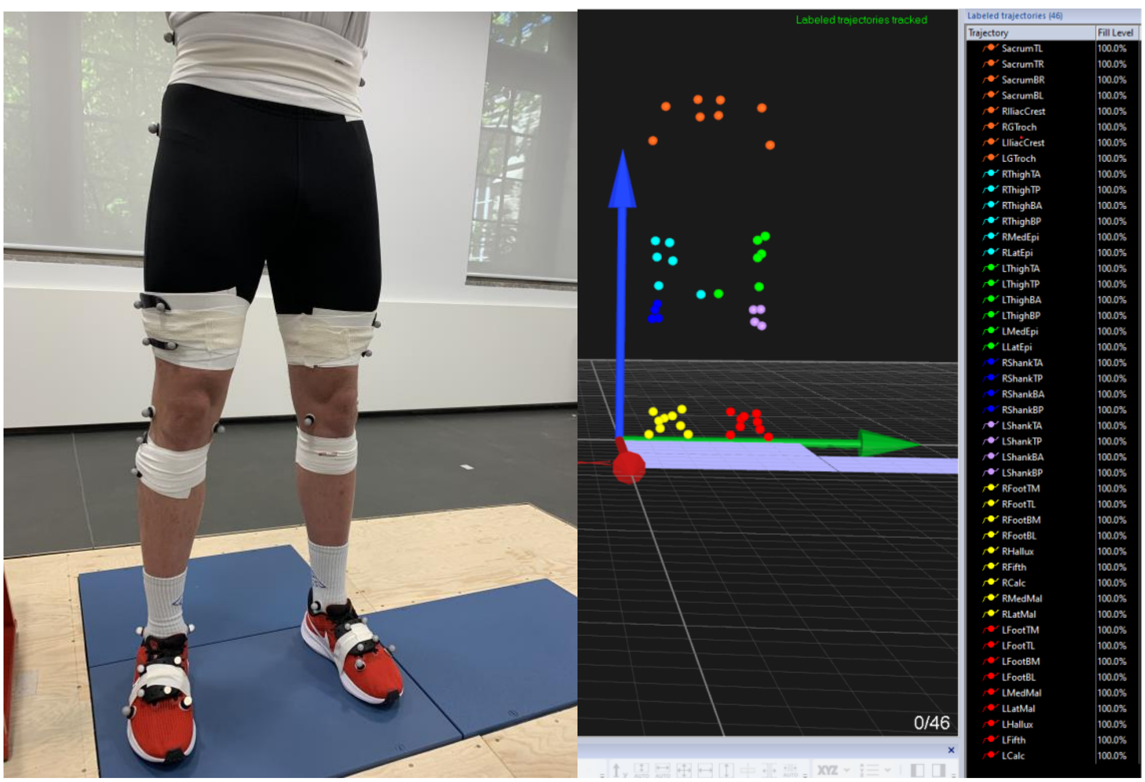

Figure 4. Marker placement and identification

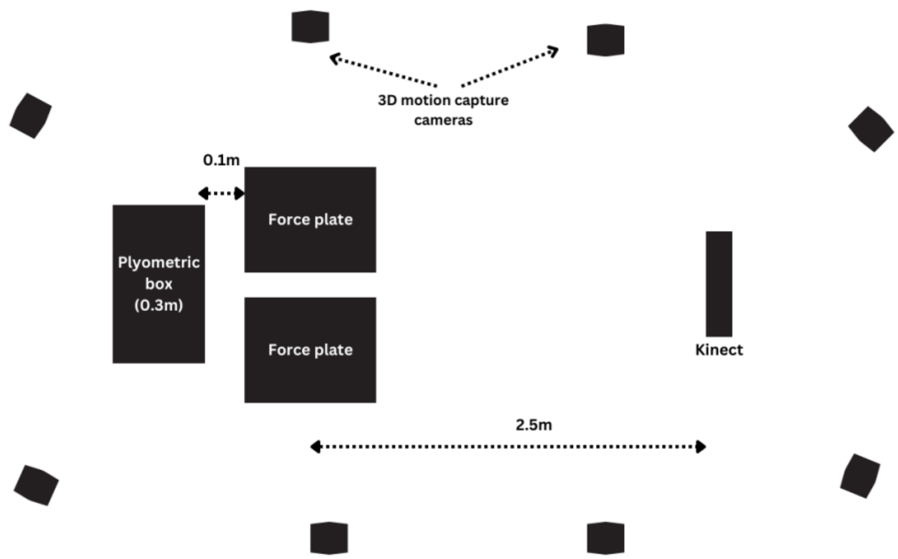

Figure 5. Laboratory setup

3.4 Procedures

Following marker placement, the participants underwent a standing calibration trial and removal of the calibration markers. The researcher verbally explained the DVJ task using standardized instructions and provided a visual demonstration. The DVJ task involved stepping off a box positioned 10cm from the force plates, landing with both feet, immediately performing a maximal vertical jump, landing with both feet simultaneously then taking 3 steps forward (Figure 6)[18]. Following the instructions and demonstration, the participant completed 3 practice DVJ trials to familiarize themselves with the task. The Microsoft Kinect and 3D motion capture system simultaneously recorded 5 DVJ trials per participant. Any trials not successfully captured by both measurement methods were repeated. A successful trial for the Kinect was where the ACL Gold software identified all points of interest and there was no visibly evident error such as the feet being airborne, or the participant walking forward in the video frames. A successful 3D motion capture trial tracked all marker positions throughout the task, and both feet made full contact with separate force plates.

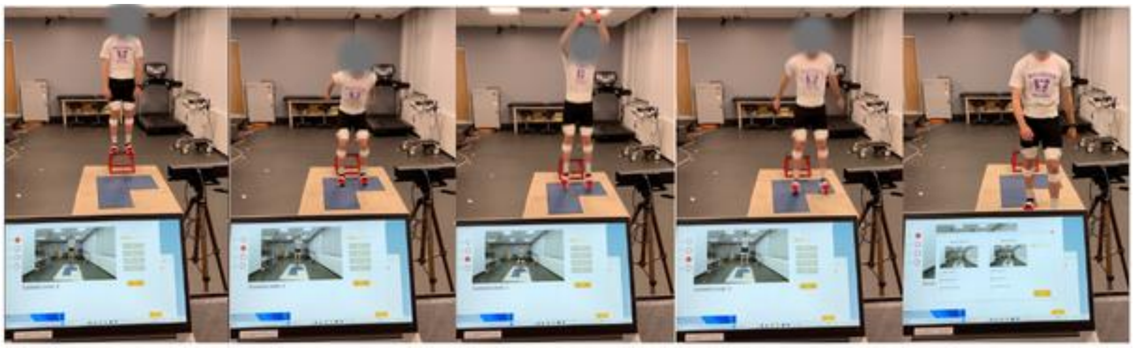

Figure 6. Drop Vertical Jump (DVJ) sequence

3.5 Data Reduction

Microsoft Kinect data were analyzed using ACL Gold software[63,106] (Figure 7), which measures (1) the KASR, which is the ratio of the distance between the knees compared to the distance between the ankles according to the equation: $KASR = \frac{Rknee\ position - Lknee\ position}{Rankle\ position - Lankle\ position}$ (Figure 8), and (2) the "knee valgus angle", or knee abduction angles, defined as the angle of the shank with reference to the thigh in the frontal plane, at initial foot contact with the ground and at peak knee flexion during landing[18] (Figure 9). The Kinect identified initial contact as the frame where the ankle joint center velocity decreases and peak knee flexion as the frame where the hip joint centers and base of the spine locations were lowest with reference to the floor[42]. The thigh segment was defined as the length between the hip and knee joints, the shank as the length between the knee and ankle. Joint centers were identified as a function of information from the depth sensor and 2D camera. The depth sensor used infrared beams and a time-of-flight sensor to gauge the distance between the Kinect and objects in the camera field[60,61]. Information from the depth sensor and the camera were input into a decision tree matrix that estimates which pixels correspond with joint center locations[60,62]. Motion capture analyses typically use the average of trials rather than individual trials to quantify habitual movement patterns. Therefore, the average KASR scores and knee abduction angles were calculated for each participant from the 5 successful trials. Similarly, the average knee abduction angles were calculated for each leg separately. No difference was found between right and left knee abduction angles, and the right leg values were used for analysis.

Model construction and data analysis from the 3D motion capture system were completed using Visual 3D (C-Motion). Marker trajectories and force plate data was lowpass filtered at 12Hz. Knee abduction angles and KASR values were extracted at initial contact and peak knee flexion during each landing. Initial contact, or 0% of landing, was defined as the frame where pressure on the force plates exceeded 20N, 100% landing was defined as the frame where pressure on the force plates fell below 20N. Peak knee flexion was determined by the sagittal plane angle between the thigh and shank segments. Knee and ankle joint centers were used to determine KASR according to the equation $KASR = \frac{Rknee\ position - Lknee\ position}{Rankle\ position - Lankle\ position}$. A KASR score of 1.0 indicates the knees are directly superior to the ankles, a score less than 1.0 indicates knee position medial to the ankles, and greater than 1 indicates knee position lateral to the ankles[18]. The knee abduction angle was defined using Euler/Cardan angles (XYZ rotation sequence) as motion of the tibia relative to the femur in the frontal plane. The hip joint center was estimated as one quarter the distance between greater trochanters. The knee joint center was defined as the midway point between the lateral and medial femoral epicondyle markers. The ankle joint center was defined by the halfway point between the lateral and medial malleoli markers.

The thigh segment was defined as distance between the hip and knee joint centers. The shank segment was defined as the length between the knee and ankle joint centers.

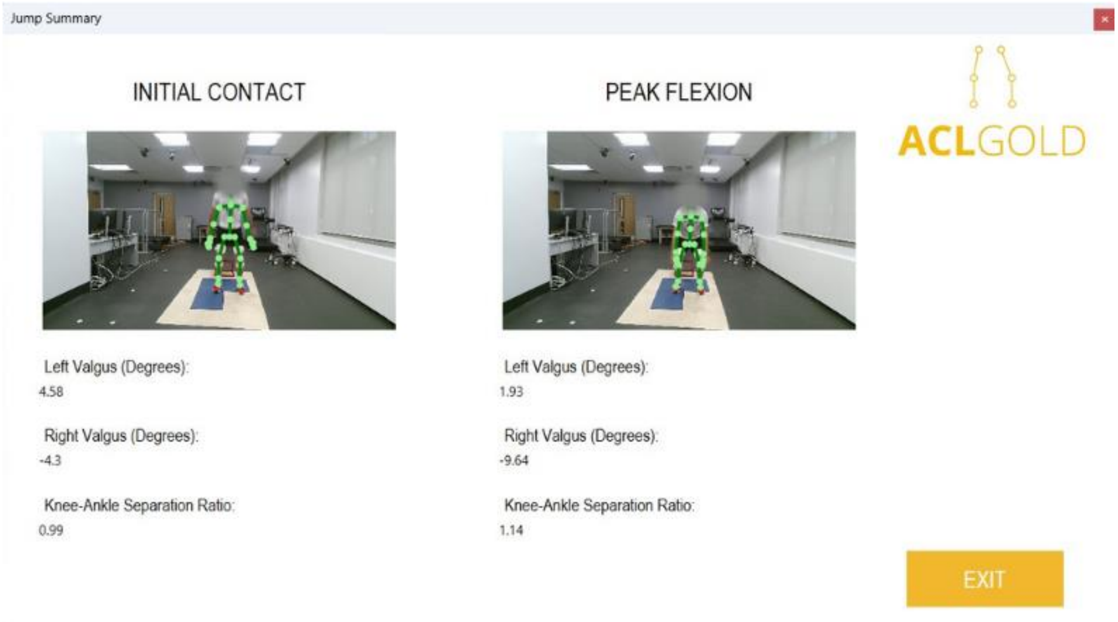

Figure 7. Kinect with ACL Gold output

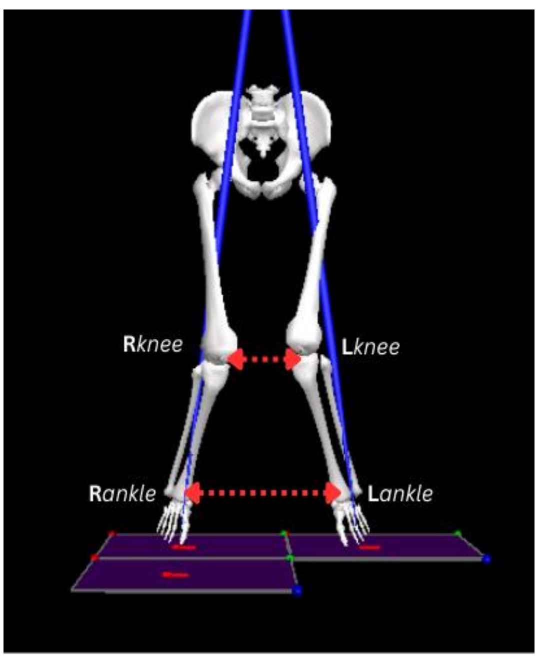

Figure 8. Knee ankle separation ratio

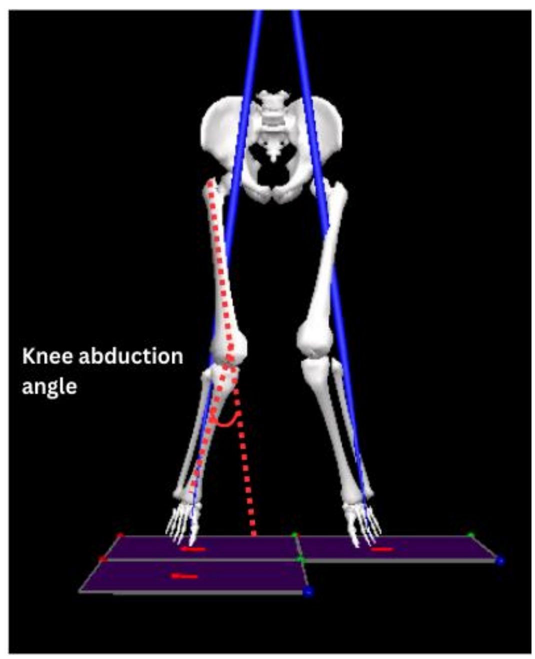

Figure 9. Knee abduction angle

3.6 Statistical analysis

Mean and standard deviation were calculated for all demographic data (Table 1). All data were inspected for normality using the Shapiro-Wilk test. Level of agreement between the Microsoft Kinect and 3D motion capture was estimated using intraclass correlation coefficient ($ICC_{,2k}$) within each group (Table 3). 95% confidence intervals were calculated for each $ICC_{2,k}$. Established guidelines for reporting agreement consider ICC values <0.5 as poor, 0.50-0.75 as moderate, 0.76-0.90 as good and 0.90-1.0 as excellent[107]. Knee ankle separation ratio and knee abduction angle measurements were averaged per subject and separated by group and method (Table 3.) The influence of measurement method, sex and BMI were evaluated using a 2 (sex) by 2 (BMI) by 2 (measurement method) ANOVA (α=0.05) for KASR and knee abduction angles at initial contact and peak flexion. Post hoc comparisons evaluated significant interactions using independent samples t-tests and Bonferroni adjustment. Scatterplots were created to further visualize agreement between methods for each group (Figures 11-14). Bland-Altman plots were generated to visualize systematic bias (Figures 15-18).

Chapter 4

4 Results

4.1 Participant information

Subject demographics by group are reported in Table 1. The male and female groups with high body size had greater mass and BMI than their matched counterparts as expected, and the males were taller than females.

One outlier was identified in the high BMI male group in knee abduction at peak knee flexion during 3D motion capture analyses (Figure 12). After visual inspection of the biomechanical model, it was concluded that there was no error in data collection and the values were biologically plausible based on previously reported data[95,108]. As such the individual was retained for all analyses that evaluated agreement between methods. Group comparisons were conducted with and without the outlier. As the interpretation did not differ when the outlier was excluded (Appendix 5), we chose to retain the individual as per the intended analyses.

Table 1. Demographic information

	Low BMI female (n=10)	**High BMI female (n=10)**	**Low BMI male (n=10)**	**High BMI male (n=10)**
Age (years)	20.9 ±2.6	20.8±2.0	21.6 ±1.3	21.0 ±1.5
Limb dominance (n)	R 8/L 2	R 10/L 0	R 9/L 1	R 7/L 3
Height (m)	1.70 ±0.07	1.67 ±0.10	1.81 ±0.07	1.85 ±0.07
Mass (kg)	65.24 ±4.85	97.89 ±17.75	77.31 ±7.42	118.30 ±15.78
BMI (kg/m^2)	22.52 ±1.07	34.99 ±4.40	23.51 ±1.38	34.56 ±3.18

Note: Low BMI is 18.5-24.9 kg/m^2**,** high BMI indicates BMI >30 kg/m^2

4.2 Agreement between methods

Agreement between the Kinect and 3D motion capture for KASR was moderate-to-good at initial contact (ICC Range: 0.667 – 0.861, Table 2) and peak knee flexion (ICC Range: 0.766 – 0.882, Table 2). Scatterplots also indicated weak to strong linear associations for KASR measurements at initial contact (R^2 Range: 0.044 – 0.593, Figure 10) and at peak knee flexion (R^2 Range: 0.107 – 0.874, Figure 11).

Knee abduction angles showed low to moderate agreement between Kinect and 3D motion capture for all groups at initial contact (ICC Range: 0.128 – 0.575, Table 2) and peak knee flexion (ICC Range: 0.315 – 0.760, Table 2). Scatterplots also indicated weak to moderate linear associations for knee abduction measurements at initial contact (R^2 Range: 0.011 – 0.163, Figure 12) and at peak knee flexion (R^2 Range: 0.088 – 0.412, Figure 13).

Bland-Altman plots illustrate the average differences between methods per sex and body size group for KASR and knee abduction angles at initial contact and peak knee flexion (Figures 14-17). Average differences between the Kinect and 3D motion capture greater than zero indicates the Kinect overestimated KASR at peak flexion.

Table 2 Agreement between methods (ICC_{2k} [95% Confidence Interval])

	Low BMI Female	**Low BMI Male**	**High BMI Female**	**High BMI Male**
KASR at IC	.861 [.442, .966]	.667 [-.340, .917]	.831 [.319, .958]	.728 [-.095, .932]
KASR at PKF	.766 [.059, .942]	.882 [.525, .971]	.806 [.218, .952]	.805 [.217, .952]
Knee abduction angle at IC	.360 [-1.576, .841]	.172 [-2.333, .794]	.128 [-2.509, .783]	.575 [-.713, .263]
Knee abduction angle at PKF	.315 [-1.759, .830]	.582 [-.681, .896]	.533 [-.879, .884]	.760 [.033, .940]

Note: IC = initial contact, PKF = peak knee flexion

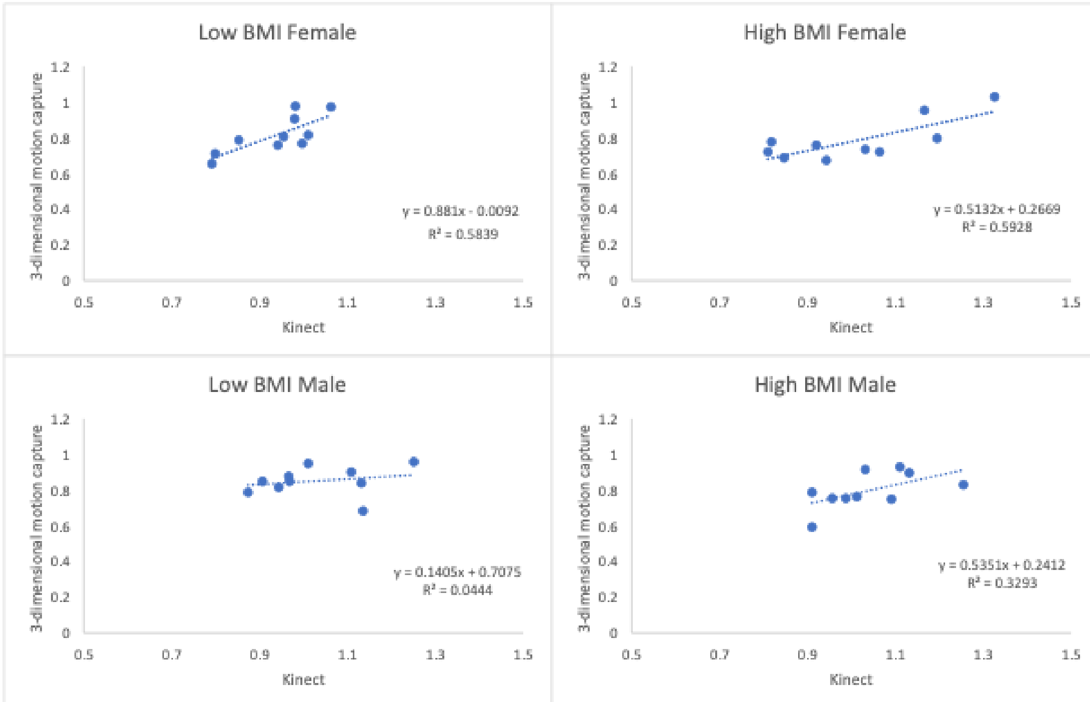

Figure 10. Average KASR at initial contact per sex and BMI group.
Note: KASR =1 indicates neutral alignment of knees relative to ankles

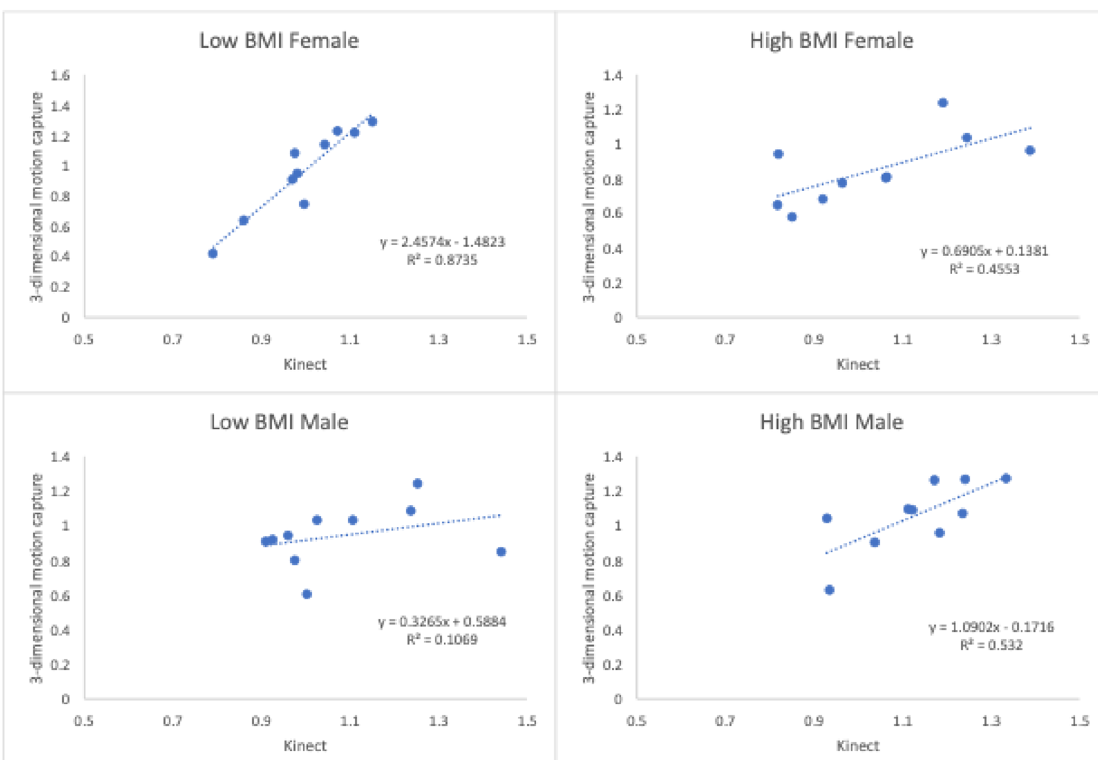

Figure 11. Average KASR at peak knee flexion per sex and BMI group
Note: KASR = 1 indicates neutral alignment of the knees relative to the ankles

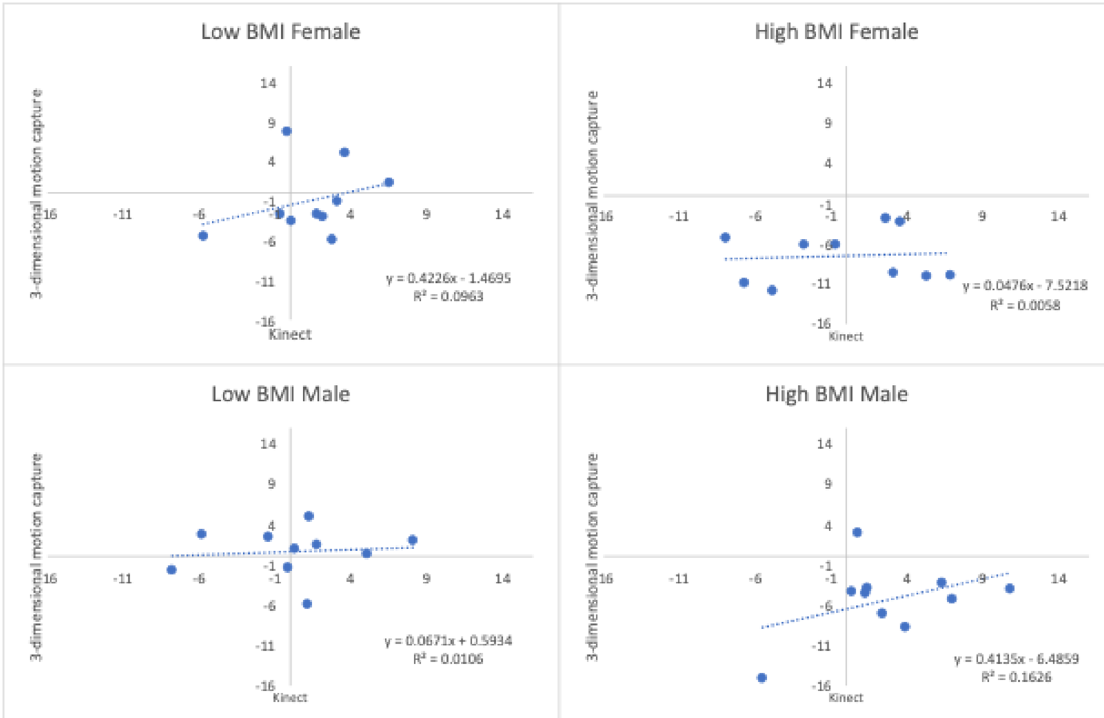

Figure 12. Average knee abduction angles at initial contact per sex and BMI group
Note: Knee abduction angle=0 indicates neutral knee alignment

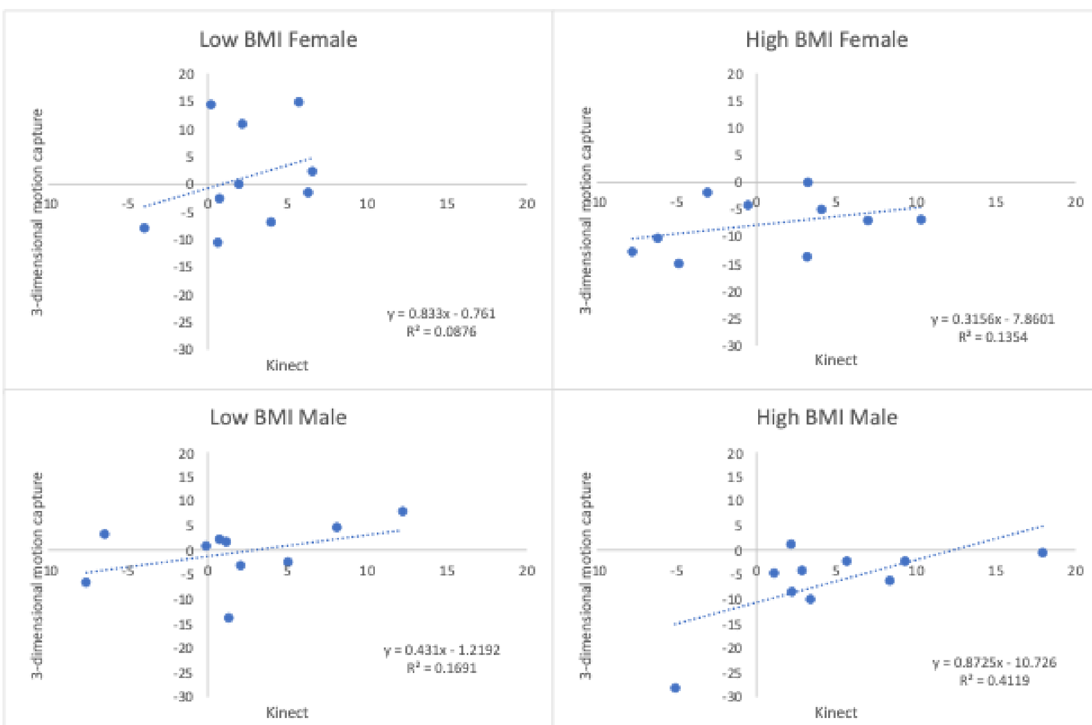

Figure 13. Average knee abduction angles at peak knee flexion per sex and BMI group
Note: Knee abduction angle=0 indicates neutral knee alignment

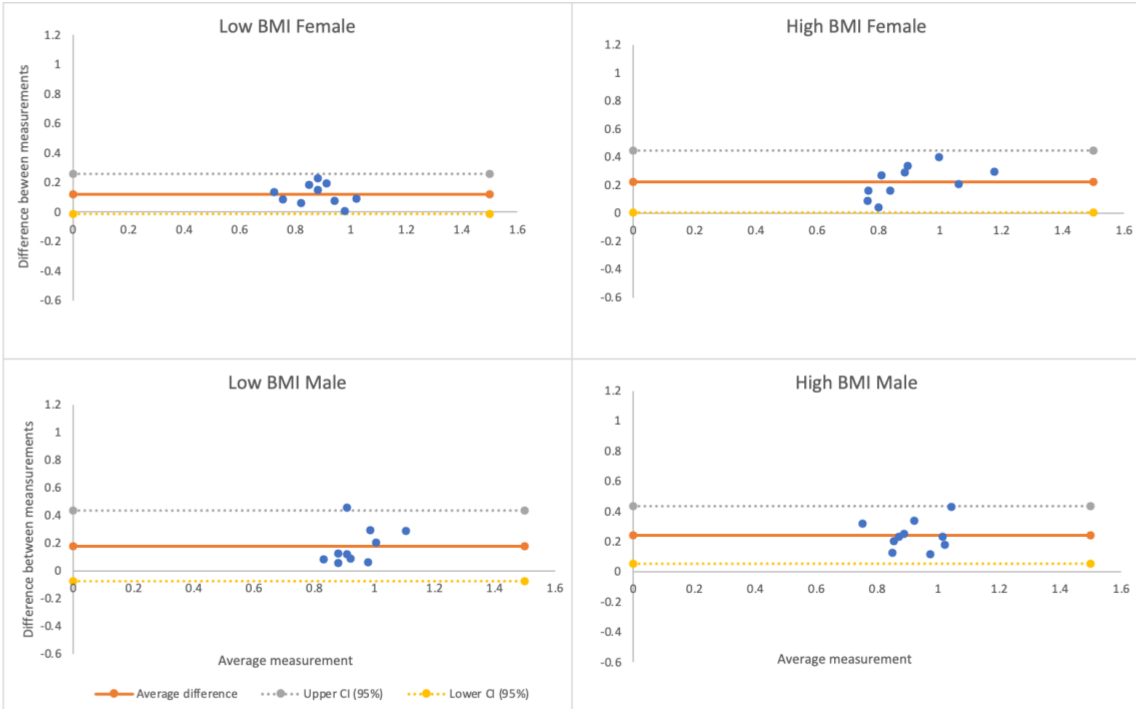

Figure 14. Bland-Altman plots for KASR at initial contact

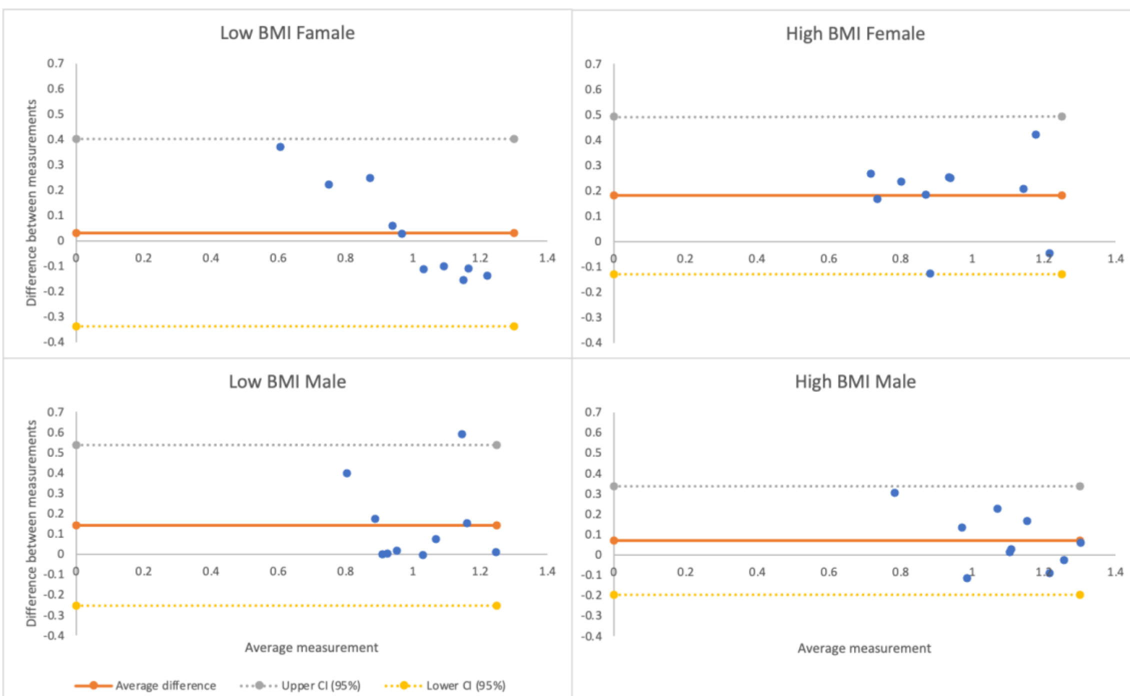

Figure 15. Bland-Altman plots for KASR at peak knee flexion

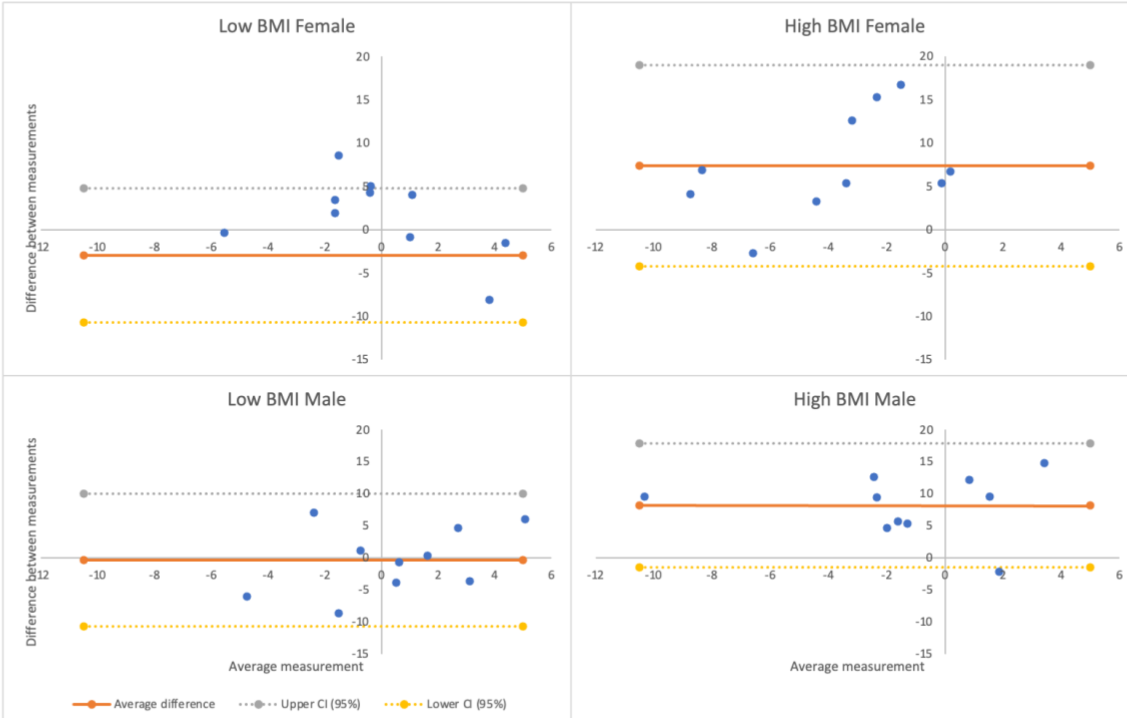

Figure 16. Bland Altman plots for knee abduction angle at initial contact

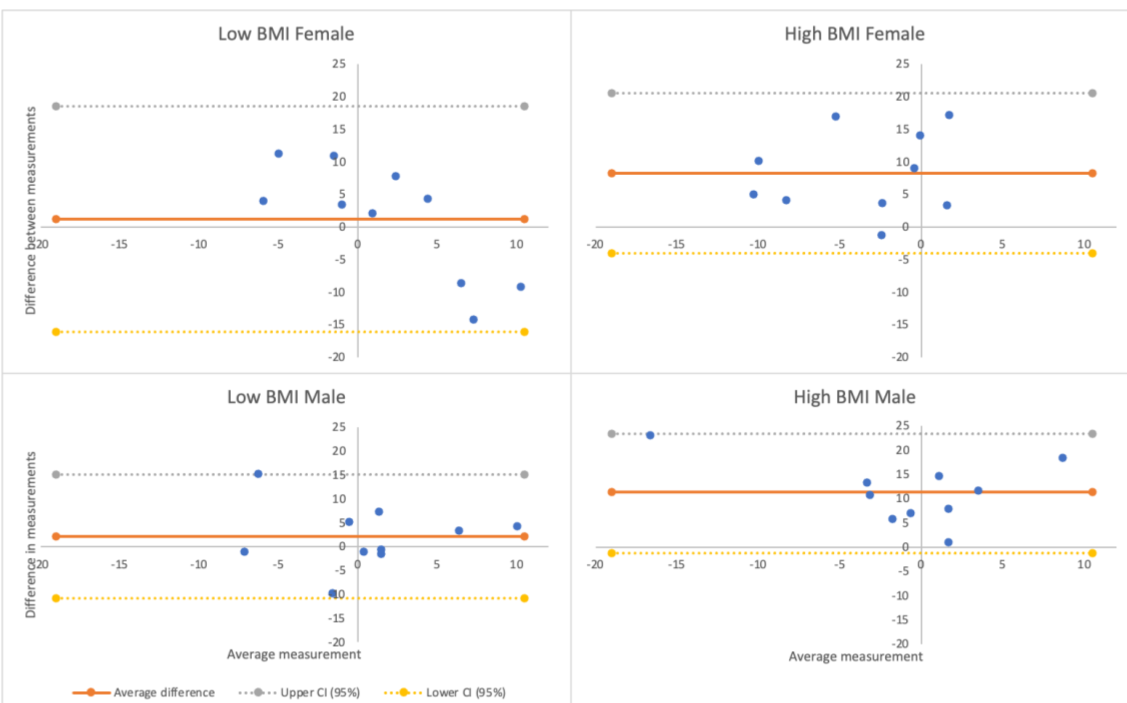

Figure 17. Bland-Altman plots for knee abduction angle at peak knee flexion

4.3 Group comparisons

Ensemble average waveforms for knee abduction angles during the DVJ were plotted per group for visualization purposes (Figure 18). Mean KASR and knee abduction angles at initial contact and peak knee flexion by group per method are reported in Table 3.

Table 3. Mean KASR and knee abduction measurements

	Low BMI Female (n=10)		**Low BMI Male (n=10)**		**High BMI Female (n=10)**		**High BMI Male (n=10)**	
	Kinect	Motion Capture	Kinect	Motion Capture	Kinect	Motion Capture	Kinect	Motion Capture
KASR IC (0-2.0)	.938 (.092)	.817 (.106)	1.03 (.121)	.872 (0.055)	1.012 (.175)	.786 (.117)	1.04 (.109)	.799 (.101)
KASR PKF (0-2.0)	.995 (.109)	.964 (.286)	1.044 (.120)	.965 (.136)	1.032 (.194)	.851 (.199)	1.130 (.132)	1.060 (.197)
Knee Abduction IC (º)	1.332 (3.270)	-.907 (4.451)	.244 (4.634)	.610 (3.022)	-.170 (5.210)	-7.53 (3.250)	2.842 (4.462)	-5.31 (4.576)
Knee Abduction PKF (º)	2.416 (3.283)	1.251 (9.241)	1.630 (5.972)	-5.167 (6.259)	.524 (5.980)	-7.695 (5.125)	4.781 (6.130)	-6.555 (8.333)
DVJ attempts (n)	7.8 (2.616)		8.8 (4.264)		7.8 (2.573)		13 (3.590)	

Note: IC = initial contact, PKF = peak knee flexion**,** Knee abduction angles were measured in degrees (- indicates more abducted)**.** DVJ attempts = average number of attempts required to collect 5 successful trials.

4.3.1 KASR at initial contact

The sex by BMI by method interaction was not significant for KASR at initial contact ($p>0.05$, Table 4). There were also no significant interaction effects between sex and method, or sex and BMI (all $p>0.05$, Table 4). There was a significant BMI by method interaction ($F_{1,36}=10.022$, $p=0.003$, partial eta square=0.218, Table 4) and no main effect of sex was observed ($p=0.162$). When collapsed across sex, post hoc analyses indicated a greater difference between methods in the high BMI group (mean difference=0.234, 95% CI: 0.191, 0.227) than in the low BMI group (mean difference=0.139, 95% CI: 0.096, 0.182). Finally, there was a significant main effect of method ($p<0.001$, Table 4) indicating a lower KASR at initial contact when measured with 3D motion capture compared with the Kinect (mean difference=-0.191, 95% CI: -0.225, -0.158).

4.3.2 KASR at peak knee flexion

There were no significant three-way or two-way interaction effects at peak knee flexion (all $p>0.05$, Table 4). There was a significant main effect of method on KASR measurement at peak flexion ($p<0.001$, Table 4) indicating a lower KASR when measured using 3D motion capture compared to the Kinect (mean difference=-0.090, 95% CI: -0.137, -0.043).

4.3.3 Knee abduction angle at initial contact

The sex by BMI by method interaction effect was not significant for knee abduction angle at initial contact ($p>0.05$, Table 4). There were also no significant interaction effects between sex and method, or sex and BMI (all $p>0.05$, Table 4). There was a significant BMI group by method interaction for the knee abduction angle at initial contact ($F_{1,36}=18.683$, $p<0.001$, partial eta squared=0.342) with no main effect of sex ($p=0.714$, Table 4). When collapsed across sex, post hoc analyses indicated that the high BMI group had a larger knee abduction angle at initial contact compared with the low BMI group (mean difference=6.272°, 95% CI: 3.778, 8.765), but only when evaluated using 3D motion capture. Moreover, the knee abduction angle was greater when measured using 3D motion capture compared with the Kinect (mean difference = 7.757° 95% CI: 5.395, 10.118), but only in the high BMI group.

4.3.4 Knee abduction angle at peak knee flexion

The sex by BMI by method interaction effect was not significant for knee abduction angle at peak knee flexion ($p>0.05$, Table 4). There were also no significant interaction effects between sex and method, or sex and BMI (all $p>0.05$, Table 4). There was a significant BMI group by method interaction for knee abduction angle at peak knee flexion ($F_{1,36}=12.985$, $p<0.001$) with no main effect of sex ($p=0.683$, Table 4). When collapsed across sex, post hoc analyses indicated larger knee abduction angles at peak knee flexion in the high BMI group than the low BMI group (mean difference=7.492°, 95% CI: 2.733, 12.251) but only when measured using 3D motion capture.

Table 4: Method by BMI by sex interactions and main effects

	Interaction Effects								Main Effects					
	Method x sex x BMI		Method x BMI		Method x Sex		Sex x BMI		Method		Sex		BMI	
	F	P	F	P	F	P	F	P	F	P	F	P	F	P
KASR at IC	0.116	0.735	10.022	0.003*	0.800	0.377	0.684	0.414	155.315	<0.001	2.039	0.162	0.022	0.883
KASR at PKF	2.921	0.960	2.320	0.136	0.490	0.488	1.523	0.225	15.174	<0.001*	2.938	0.095	0.253	0.618
KAA at IC	1.079	0.306	7.171	0.011*	2.381	0.132	0.725	0.400	5.473	0.025*	0.137	0.714	18.683	<0.001*
KAA at PKF	0.224	0.639	12.985	<0.001*	0.827	0.369	1.326	0.257	25.730	<0.001*	0.170	0.683	3.951	0.050*

*Denotes significance at P <0.005; KASR: Knee-Ankle Separation Ratio; KAA: Knee abduction angle; IC: initial contact; PKF: peak knee flexion

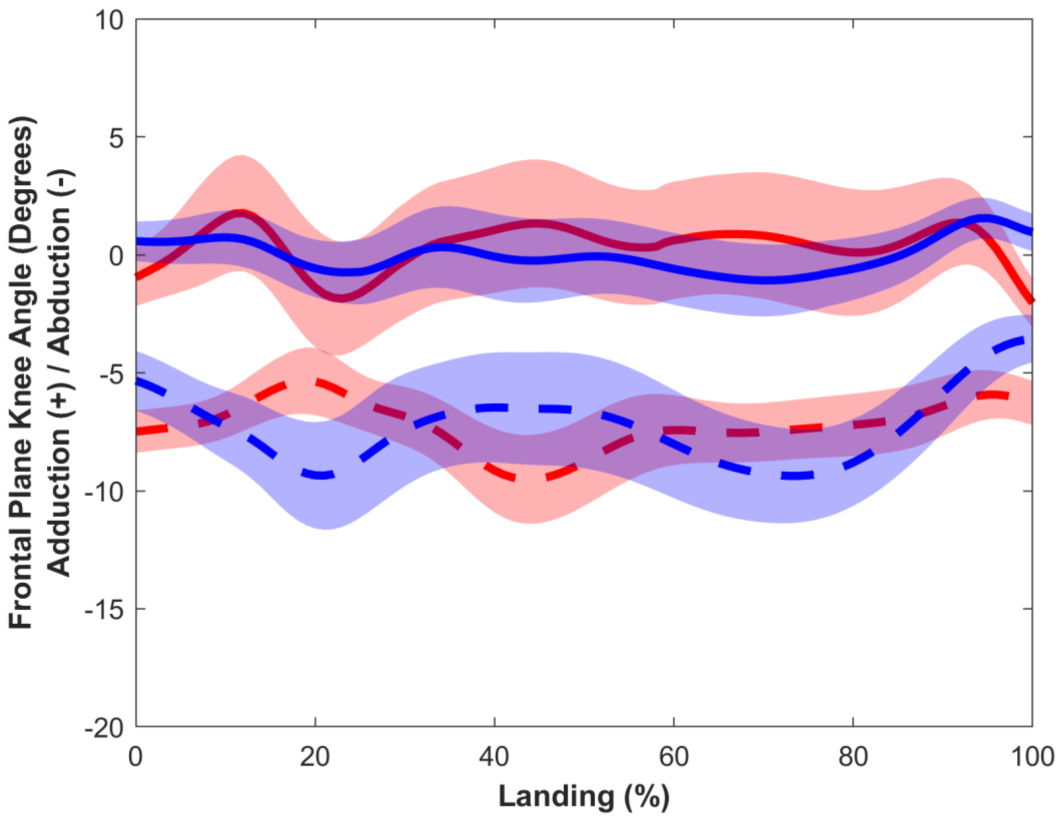

Figure 18. Knee abduction angle waveforms for 100% of landing [95% CI]

Note: Red = female, blue = male, solid line = low BMI, dotted line = high BMI

Chapter 5

5 Discussion

Dynamic knee valgus is multi-joint and multi-planar movement pattern associated with risk of ACL injury[9,30]. Knee ankle separation ratios and knee abduction angles are surrogate measurements for dynamic knee valgus[18]. This study measured KASR and knee abduction angles across sex and BMI groups to assess agreement between methods. We found moderate to good agreement between the Kinect and 3D motion capture for KASR, but low agreement for knee abduction angle measurements, and there were effects of body size on some measurements. The secondary purpose was to assess the influence of sex and BMI on landing patterns. No differences were found in the landing patterns between males and females. The high BMI group showed more medial knee position during landing than the low BMI group when collapsed across sex and measured using 3D motion capture.

5.1 Agreement between the Microsoft Kinect and 3-dimensional motion capture

5.1.1 Knee ankle separation ratio

The hypothesis of good agreement between the Kinect and 3D motion capture for measuring KASR was partially supported. The level of agreement between the Kinect and 3D motion capture for measuring KASR was moderate-to-good for all groups at initial contact and peak flexion. Knee alignment relative to the ankles in the frontal plane can be estimated by the KASR. Mizner et al. compared KASR and frontal plane projection angles from 2D measurement and knee abduction angles from 3D motion capture and found that lower KASR values were associated with greater frontal plane projection angles and knee abduction angles[16]. Sigward et al. found that frontal plane knee separation distance, normalized to intertrochanteric width, was able to predict average bilateral knee abduction angles[15]. Frontal plane projection angles and knee abduction angles are associated with medial knee displacement and risk of ACL injury[16]. Therefore, KASR may be a useful surrogate for measuring medial knee displacement and risk of ACL injury during biomechanical analyses. The level of agreement we found between the Kinect and 3D motion capture for measuring KASR shows that the Kinect could be used in clinical settings to identify movement patterns that may place individuals at increased risk of ACL injury.

The difference in the mean measurements at both initial contact and peak knee flexion between the Kinect and motion capture indicate that the Kinect is overestimating the KASR. A KASR of 1 indicates neutral alignment, with equal distance between the knees as between the ankles. The Kinect measurements are closer to 1 than the 3D motion capture, indicating that the Kinect is labeling knee alignment as more neutral when the 3D motion capture identifies medial knee positioning. Interpretation of KASR measurements from the Kinect should take into consideration the underestimation of medial knee position or recommend further biomechanical assessment for potentially higher risk individuals.

5.1.2 Knee abduction angle

The Kinect reports "knee valgus angles" for each leg, which correspond to knee abduction angles measured in the frontal plane[63]. Greater knee abduction during landing is prospectively associated with increased risk of ACL injury[9,30]. The hypothesis of good agreement between the Kinect and motion capture for measuring knee abduction angles was not supported. Agreement between methods at initial contact was low to moderate for all groups. Agreement between methods at peak knee flexion was low to moderate for all females and low BMI males, and good for high BMI males at peak knee flexion. Possible reasons for low agreement between methods include joint center estimation and identification, joint center obstruction, and the projection of hip and ankle kinematics onto frontal plane knee measurements.

5.1.2.1 Joint center estimation

Measuring the knee abduction angle requires tracking the end points of the thigh and shank segments. Thus, accuracy in identifying the hip joint centers, knee joint centers, and ankle joint centers influences the accuracy of knee abduction angle measurement[21]. Xu et al. reported differences in joint center identification of more than 60mm during standing and 100mm during a static squat between the Kinect and 3D motion capture for all lower body joint centers[21]. Errors in joint center identification, as reported by Xu et al., alter the thigh and shank segment lengths, positions, and resultant knee abduction angles[21]. Soft tissue surrounding joints and limb segments may contribute to error in joint center identification. This is supported by the interaction between BMI and method for knee abduction angle at both initial contact and peak flexion. The differences between the measurements taken by the Kinect and 3D motion capture were greater in the high BMI groups when collapsed across sex. Soft tissue distribution alters body surface area[32,38]. The Kinect used 2D video and an infrared scan of body surface area to identify joint centers[62,63]. Thus, soft tissue distribution may have contributed to the low agreement between the Kinect and 3D motion capture measurements for knee abduction angles.

5.1.2.2 Joint center obstruction

Joint center obstruction may have contributed to error in measuring knee abduction angles at peak knee flexion. At peak knee flexion, the pelvis lowers while the knees and shanks stay near vertical. Thus, at peak knee flexion, the hip joint centers may have been obstructed by the knees or shanks from a frontal plane perspective. Moreover, individuals with larger knee and hip flexion may have greater obstruction of the hip joints from the Kinect's perspective, which may reduce the ability to accurately identify the hip joint center.

5.1.2.3 Multi-planar kinematics

Furthermore, multi-plane hip and ankle kinematics may have influenced 2D measurements of frontal plane knee angles. Nagano et al. suggest that hip rotation and ankle pronation contribute to the appearance of 2D knee abduction from a frontal plane perspective[64]. As the Kinect measured only frontal plane kinematics, it did not account for apparent changes in knee position due to hip rotation or foot position. Whereas 3D motion capture was not limited to a frontal plane perspective and measured hip, knee, and ankle kinematics in all three planes. Thus, the effect of hip and ankle kinematics may have contributed to differences in measurement between the Kinect and 3D motion capture.

5.1.3 Event identification

Error in event identification may have contributed to differences between methods for both KASR and knee abduction angle measurements. The Kinect extracted KASR and knee abduction angle measurements at initial contact and peak knee flexion. Thus, the Kinect's accuracy in identifying initial contact and peak knee flexion is integral to its validity. A difference in methodology between this study and a previous study which found higher levels of agreement than this study is the temporal alignment of points of interest[22]. Stone et al. used a Microsoft software development package to temporally align the Kinect with the 3D motion capture system so kinematic measurements at initial contact were taken from identical frames determined by the position of the reflective markers identified with 3D motion capture[22]. Peak knee flexion was identified by each method separately[22]. Stone et al. reported more consistent measurements at initial contact than peak flexion, and a lower level of agreement for knee abduction angle between systems at peak knee flexion[22]. A similar validation study by Gray et al. found good to excellent agreement between the Kinect and motion capture for measuring KASR at both initial contact and peak flexion[18]. Gray et al. temporally aligned the Microsoft software development package with the motion capture system and extracted data from matched frames[18]. Gray et al. identified initial contact using the Kinect and compared the measurement to the matched motion capture frame, whereas Stone et al. used motion capture to identify initial contact and measured the matched frame from the Kinect[18,22]. Gray et al. reported the Kinect accurately identified events 95.8% of time[18]. However, the researchers determined accuracy visually rather than verifying it against force plate data or 3D motion capture [18]. Our study was not able to align initial contact, and each system identified events of interest independently. The motion capture system identified initial contact as the instance when force on the force plates exceeded 20N and peak knee flexion as the point of greatest sagittal plane knee angle occurring while the feet were in contact with the force plates. The Kinect identified initial contact as the frame where the ankle joint centers' velocity decreased, indicating contact with the ground[18]. The Kinect considered peak knee flexion as the instance where the hip joint centers and base of the spine locations were lowest relative to the floor[18,109]. Thus, the levels of agreement we found between methods may be partially due to differences in event identification.

Accurate event identification is a critical component of accurate kinematic measurements. In a clinical setting, the Kinect would not be temporally aligned with 3D motion capture. Thus, assessing agreement between methods without temporal alignment is representative of using the Kinect in a practical scenario rather than a laboratory setting.

Discrepancies in joint center locations of the ankles, hips and base of the spine may have contributed to differences in event identification between methods. Xu et al. reported left and right ankle joint position discrepancies between the Kinect and motion capture of 126 ± 44mm and 112 ± 39mm respectively in an upright position, similar the body position at initial contact[21]. The Kinect used ankle joint velocity to identify initial contact, and velocity is a function of change in position over time. Therefore, incorrect ankle joint position identification may have contributed to event identification error. Individual landing strategies include varying degrees of dorsiflexion or plantar flexion, which present different limb surfaces to the Kinect sensor. Foot positioning at initial contact may have contributed to error in ankle position identification. The Kinect identified the peak flexion frame as the point where the hip joint and base of the spine were lowest in reference to the floor. The lowest position of the hips and base of the spine is a surrogate for peak knee flexion, provided the hip and base of spine locations are accurate. Thus, error in hip and spine identification may have contributed to inaccurate identification of the peak knee flexion frame. Inaccuracy in joint center identification in a squat position, similar to peak knee flexion during landing, may be due in part to the hips and base of the spine being obstructed by the knees and shanks. Xu et al. reported left and right hip joint location discrepancies in a squat position as 109 ± 59mm and 110 ± 52mm for left and right hips respectively[21]. The difference between the location of the spine base identified by the Kinect and motion capture was 206 ± 50mm in a squat position[21]. Moreover, Xu et al. reported that their measurements were not applicable to groups with high BMI [21]. Differences in event identification may have caused kinematic measurements to be extracted from different frames and contributed to lower agreement between methods.

5.2 Practical application

5.2.1 Unsuccessful trials

During several DVJ attempts, the Kinect was unable to generate measurements. Therefore, to collect sufficient measurements, the subject completed more than the anticipated 5 DVJ attempts. The high BMI groups required more attempts to collect 5 successful trials than the low BMI groups. Repeating attempts required additional testing time and may have placed additional strain on the individual. If many additional attempts are required, fatigue and a learning effect may influence measurements. Therefore, additional rest periods may be needed when assessing participants with larger body size.

There were multiple reasons for DVJ attempts to either not be collected or to be unusable. Following some attempts, the ACL Gold software showed an error dialogue box rather than the measurement window. Error codes were “unable to identify points of interest” or “no difference between initial contact and peak flexion”. When the Kinect is “unable to identify points of interest”, it did not specify whether it could not identify initial contact, peak knee flexion, or both. However, previously measured differences in hip joint center and base of spine locations between the Kinect and motion capture during a squat indicate inaccuracy or an inability to track the landmarks used to identify peak knee flexion[21]. The error code “no difference between initial contact and peak flexion” indicated a similar inability of the Kinect to identify the relevant joint centers. Xu et al. reported average differences in knee position identification between the Kinect and 3D motion capture in standing and squatting positions as 67 ± 55mm and 118 ± 50mm respectively[21]. Thus, the Kinect may have registered no change in joint position if an individual landed with little knee flexion and knee position was underestimated at initial contact and overestimated at peak knee flexion. Trials were also discarded based on visual inspection of the images of the Kinect frames. If subjects’ feet were visibly airborne or the image showed the subject walking or in a non-landing stance, the attempt was discarded. Checking each image for visible errors placed increased demand on the user and introduced possible human error to the measurement. Failure to discard erroneous measurements may have contributed to errors in determining average knee motion during landing.

5.2.2 Camera perspective

The Kinect has a 2D camera, and a depth sensor adds a third dimension as the distance between the sensor and the object. However, the depth sensor does not allow kinematic measurement in the sagittal or transverse planes. Stiff knee landings are associated with risk of knee injury and are characterized by less knee and hip flexion than soft landings, and less kinetic energy absorbed by the muscular system[110]. Thus, more kinetic energy is transmitted to the inert stabilizing structures such as the ACL during stiff knee landings[110]. By not measuring sagittal knee angles, the Kinect may miss important kinematic measurements related to ACL injury, particularly in individuals who may be at greater risk of ACL injury due to stiff kneed landing patterns.

5.2.3 Kinect output interpretation

The output from the ACL Gold Software provides the KASR and knee abduction angle outcome measures per event and the frame number they were extracted from. Understanding mechanisms of ACL injury and landing patterns associated with risk of ACL injury may guide interpreting values at initial contact or peak knee flexion. Previous research on the timing of ACL injury during landing suggests that rupture typically occurs closely following initial contact and may be associated with knee flexion angles during landing[10,111–113]. A review of research on factors contributing to ACL injury reports conflicting results on how knee flexion angle influences ACL injury[113]. The review suggests that different factors associated with risk of ACL injury may be present at different degrees of knee flexion[113]. This is supported by Krosshaug et al. who suggested that both high and low degrees of knee flexion during landing may be associated with risk of ACL injury[10]. Hewett et al. also reported greater knee abduction angles at both initial contact and peak knee flexion in athletes who subsequently injured their ACLs compared to uninjured athletes[14]. As both greater and lower degrees of knee flexion have been associated with ACL injury risk, evaluating knee kinematics at initial contact and peak knee flexion may be related to different factors involved in risk of ACL injury.

Initial contact occurs prior to any force absorption during landing. Therefore, interpreting Kinect measurements at initial contact may be reflective of static alignment rather than dynamic motion. Static alignment of the lower limb is influenced by non-modifiable factors such as Q-angle, pelvic shape, and tibial plateau angles[34,68]. Knee position at initial contact may also be important for individuals with stiff knee landing strategies. Accurate measurement of medial knee positioning in extension is important for those at increased risk of injury due to stiff knee landing strategies involving less knee flexion.

Conversely, peak knee flexion occurs after the initial loading phase of landing. The initial loading phase requires strength and neuromuscular control of lower body musculature for force absorption and maintaining a neutral knee position[113,114]. McCurdy et al. found that greater hip abductor strength was associated with lower mean valgus angles during both single and double limb landings[114]. Therefore, interpreting Kinect measurements at peak knee flexion may be more related neuromuscular control than knee position at initial contact[51].

Neuromuscular control is a modifiable factor associated with risk of ACL injury[12,27]. Pre- and post- intervention biomechanical analysis of jumping tasks have shown effects of neuromuscular training on knee abduction angle and normalized knee separation distances[79,115]. Furthermore, neuromuscular training has been associated with decreased incidence of ACL injury in soccer, volleyball, and basketball athletes[80]. Good agreement between the Kinect and 3D motion capture for KASR at peak knee flexion across all groups indicates potential for using the Kinect to identify individuals who may benefit from a neuromuscular intervention. Further research associating KASR values at both initial contact and peak knee flexion with risk of injury may provide clinical benchmarks, or cut-off levels for determining individuals who are at greater risk of ACL injury. The influence of BMI on measurement method and KASR during landing, and the previously established differences in landing patterns and incidence of ACL injury between sexes indicates that prospective clinical benchmarks may vary by sex and BMI[25,66,78]. Furthermore, the Kinect may be used over the course of a neuromuscular exercise intervention to quantify changes in landing patterns. Thus, the Kinect could objectively measure the efficacy of interventions.

The association between KASR and frontal plane knee angles have been established. However, the direct association between KASR and risk of ACL injury may strengthen the use of KASR in injury risk screening[15,16]. Stone et al. calculated the difference between KASR at initial contact and peak flexion to measure medial knee movement during landing[22]. As the feet are in contact with the ground, the assumption was the ankle distance remains constant between initial contact and peak flexion. With that assumption, any difference in KASR values between initial contact and peak flexion are attributable to medial knee displacement. Common mechanisms of ACL injury involve dynamic and uncontrolled movement into positions which place excessive tension on the ACL, including medial knee displacement[9,111]. Further research on the association between KASR during landing and risk of ACL injury is needed to establish criteria for screening protocols.

5.2.4 Accessibility

Benefits to using the Kinect compared to 3D motion capture for risk of injury screening include lower cost of technology, portability and low levels of training needed for use. Motion capture analyses require outfitting the subject with reflective markers, completing calibration trials during data collection, and extensive time checking marker trajectories in post processing. The Kinect is efficient as there is no need for reflective markers, calibration trials, or post-processing per trial. Cost and simplicity make the Kinect a user-friendly alternative to motion capture. However, the Kinect was only able to measure the KASR with good agreement with 3D motion capture, and precise quantifications of knee abduction should be conducted using 3D motion capture.

5.3 Landing patterns between groups

5.3.1 Body Mass Index

The hypothesis that females and participants in the high BMI groups would land with greater medial knee positioning was partially supported. Knee ankle separation ratio was not significantly different between BMI groups when collapsed across sex and measured by either method. Participants with high BMI had greater knee abduction than participants with low BMI, but this finding was only evident when evaluated with 3D motion capture. Therefore, the Kinect may not be able to measure knee abduction and detect differences between BMI groups. This is supported by low to moderate agreement between 3D motion capture and the Kinect for measuring knee abduction angle at initial contact or peak knee flexion for all sex and BMI groups. Inaccuracy in measuring the knee abduction angles, as shown by low levels of agreement, is reflected in the Kinect's inability to distinguish kinematic differences between body size groups.

When measured by 3D motion capture and collapsed across sex, participants in the high BMI groups landed with more knee abduction than their low BMI counterparts. Furthermore, visualization of the ensemble waveforms of the knee abduction angles (Figure 18) illustrate greater knee abduction in the high BMI groups throughout the entire landing phase. KASR measurements were not different between BMI groups when measured by 3D motion capture or the Kinect. The difference between the high and low BMI groups for knee abduction angles but not KASR when measured by 3D motion capture suggests that the knee abduction angle may be a more sensitive kinematic measurement of inward knee motion during landing. Therefore, the KASR should be considered a surrogate for knee abduction angle and may not be accurate in persons with larger body size.

The combination of tension on the ACL due to an abducted knee position and greater axial load due to body size places high BMI individuals at increased risk of ACL injury[28]. High BMI is also associated with increased likelihood of concomitant injury such as meniscal tears or cartilage lesions accompanying ACL injuries[83]. Concomitant cartilage injury may require additional intervention during ACL reconstruction surgery and is associated with poorer long term outcomes following reconstruction[5,6,49,83]. Poor long term outcomes following ACL injury include joint degeneration and the development of post-traumatic osteoarthritis (PTOA) in the patellofemoral and tibiofemoral joints[5,36,49,88]. ACL injury disrupts knee joint stability and alters joint loading patterns in both reconstructed and ACL deficient patients[88]. Altered loading patterns and greater compressive forces attributed to greater BMI may contribute to deterioration of articular cartilage which hasten the onset of PTOA[36,49,88]. Therefore, screening individuals with a high BMI for movement patterns that place them at increased risk for ACL injury may identify individuals who would benefit from further biomechanical analyses or neuromuscular training interventions.

Although there was no significant difference in KASR between high and low BMI groups when measured by either method, BMI influenced the measurement methods at initial contact. There was a greater difference between methods in the high BMI group compared to the low BMI group. The Kinect used a scan of body surface area to identify joint center locations, thus larger body size may have contributed to error in the Kinect's measurements. Clinical applications for the Kinect should take into consideration the contribution of BMI to measurement error when interpreting measurements of individuals with greater BMI.

There was no BMI by method interaction between KASR measurements at peak flexion. The absence of an effect of BMI at peak knee flexion may reflect different landing strategies between BMI groups that influence the measurement methods[116,117]. Previous research has shown that individuals with obesity land with less knee flexion than individuals without obesity[116,117]. Thus, at peak knee flexion, the hip center and base of the spine may not have been obstructed from the Kinect camera, leading to accurate identification of the peak knee flexion event and good agreement between the Kinect and motion capture.

5.3.2 Sex

The results from this study did not identify a difference in landing technique between males and females. However, previous research has identified sex-related differences in landing patterns and ACL injury incidence[24,25,66,78]. Females experience increased incidence rates of non-contact ACL injury compared to males[25,66]. Along with knee abduction, hip adduction and internal rotation, tibial rotation, and foot positioning during landing may contribute to the greater incidence of ACL injury in females than males[11]. This study only assessed frontal plane knee kinematics, and only during a double-limb landing task. Thus, contributions from other joints and planes of motion that contribute to ACL injury risk or single-limb tasks were not considered. Despite our results not showing a significant difference between sexes, measuring medial knee motion during landing may be a method of identifying females who would benefit from further analyses and preventative intervention. Good agreement between the Kinect and motion capture for measuring KASR for females in high and low BMI groups suggests that the Kinect may be a useful tool for screening females for high-risk landing patterns. Referring females who demonstrate medial knee positioning during landing for preventative neuromuscular intervention may contribute to closing the female to male gap in ACL injury incidence.

5.4 Limitations and strengths

5.4.1 Limitations

There are limitations to consider when interpreting the results of this study. Motion capture is considered the gold-standard for motion analysis. However, there is an expected amount of error due to soft tissue, or motion artifact. Motion artifact may occur due to motion of the surface markers in relation to the corresponding rigid segment, or from motion of the markers in reference to each other. The error from soft tissue artifact in motion capture is higher in high BMI groups, particularly around the pelvis and thigh[82,118]. Thus, error due to soft tissue artifact may be greater in the high BMI groups in this study. This study mitigated the effect of soft tissue artifact through rigid marker clusters affixed to the thighs, shanks, and sacrum. Rigid clusters provide at least 3 non-colinear markers on each limb segment that do not move in relation to each other. Thus, kinematic analyses are not influenced by the movement of markers in relation to one-another.

Secondly, real world landing scenarios include unilateral or staggered landings that load the knees asymmetrically. Within participant landing technique may differ between double and single limb landings. Landing during a sport scenario may also be accompanied by perturbations or cognitive distractions. Therefore, the DVJ task may not reflect realistic motions outside the laboratory environment. However, that does not eliminate the usefulness of bilateral measurement in a controlled environment for measuring neuromuscular function. While the DVJ may not reflect habitual motions, it may still identify individuals who would benefit from a preventative intervention or further biomechanical analyses.

5.4.2 Strengths

A strength of this study was the comparison of average measurements per participant between methods. In biomechanical analyses, a single trial may not be representative of an individual's typical movement pattern[119]. Previous analysis of a repeated hop test suggests that 4.17 ±2.12 trials are needed to achieve a representative average of frontal plane knee kinematics during landing[119]. Previous validation studies comparing the Kinect and 3D motion capture compared individual trials between methods[18,22]. Conversely, this study compared the averages from 5 measurements per participant to represent typical movement patterns. Between group comparisons also evaluated the effects of sex, BMI, and measurement method on knee abduction during landing.

Another strength of this study was the use of standardized footwear and compression shorts. Foot orientation and pronation are related to dynamic knee valgus during landing[11]. Thus, differing levels of support or cushioning in footwear may contribute to altered knee kinematics during landing. Different shapes and sizes of footwear may also alter ankle joint center identification, which could have influenced KASR measurements and event identification. Standardized footwear controlled for kinematic differences due to cushioning or support and for different joint center identification due to the shape of the footwear. Loose fitted clothing contributes to error in motion capture when markers placed on clothing move in relation to the segment they represent. Loose fitted clothing may also influence the visual and depth information gathered by the Kinect. We controlled for error due to clothing by providing compression shorts.

This study excluded the 25.0-29.9kg/m^2 BMI range. Excluding a range between the low and high BMI groups is both a strength and limitation of this study. Excluding a middle range ensured a difference in mean group BMI between low and high BMI groups since there can be considerable variation in body composition between individuals of similar BMI. Body mass index is an indicator of body size but does not account for body composition. However, adipose and muscle tissue both contribute to error in marker placement and soft tissue artifact in 3D motion capture[118,120]. Furthermore, adipose and muscle tissue both influence body surface area, which may contribute to error in the

Kinect joint center estimations[21]. Thus, body size is an important consideration when evaluating validity of motion capture methods.

Chapter 6

6 Conclusion

The Microsoft Kinect with ACL Gold software had moderate-to-good agreement with 3D motion capture for measuring KASR at initial contact and peak knee flexion for all sex and body size groups. This supports the use of the Kinect as a measurement tool for measuring KASR during landing for screening for knee injury risk. Conversely, the Kinect had low-to-moderate agreement for measuring knee abduction angles.

Body size, but not sex, influenced agreement between the Kinect and 3D motion capture, particularly for measuring knee abduction angles. The Kinect underestimated the knee abduction angles at initial contact and peak knee flexion compared to the 3D motion capture for the high BMI groups regardless of sex.

The high BMI group, regardless of sex, had greater knee abduction angles during landing than the low BMI groups, but only when measured with 3D motion capture. This supports our finding of low agreement between the Kinect and 3D motion capture for measuring knee abduction angles. The Kinect cannot accurately measure knee abduction angles and is not sensitive enough to identify differing landing patterns between body size groups.

Further research into the association between KASR and ACL injury incidence may contribute to the development of clinical benchmarks. Establishing a cut off for hazardous knee motion during landing may guide clinical decision making. Furthermore, repeated measures studies to assess the reliability of the Kinect over multiple testing sessions may guide the use of the Kinect for tracking changes in landing patterns over the course of clinical interventions or rehabilitation protocols.

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Appendices

Appendix 1. Ethics approval letter

Date: 2 September 2022

To: Dr Derek Pamukoff

Project ID: 120801

Review Reference: 2022-120801-70412

Study Title: The influence of sex and body size on the validity of the Microsoft Kinect for measuring knee motion during landing

Application Type: HSREB Initial Application

Review Type: Delegated

Full Board Reporting Date: 13/Sept/2022

Date Approval Issued: 02/Sep/2022 14:25

REB Approval Expiry Date: 02/Sep/2023

Dear Dr Derek Pamukoff

The Western University Health Science Research Ethics Board (HSREB) has reviewed and approved the above mentioned study as described in the WREM application form, as of the HSREB Initial Approval Date noted above. This research study is to be conducted by the investigator noted above. **All other required institutional approvals and mandated training must also be obtained prior to the conduct of the study**.

Documents Approved:

Document Name	Document Type	Document Date
data collection sheet	Other Data Collection Instruments	02/Sept/2022
Research protocol	Protocol	31/Aug/2022
recruitment email	Recruitment Materials	31/Aug/2022
Recruitment poster	Recruitment Materials	31/Aug/2022
recruitment slide	Recruitment Materials	31/Aug/2022
recruitment email	Email Script	31/Aug/2022
follow up email	Email Script	31/Aug/2022
Letter of Consent	Written Consent/Assent	31/Aug/2022
Medical questionnaire	Written Consent/Assent	31/Aug/2022

Documents Acknowledged:

Document Name	Document Type	Document Date
Thesis Budget	Study budget	14/Jun/2022

No deviations from, or changes to, the protocol or WREM application should be initiated without prior written approval of an appropriate amendment from Western HSREB , except when necessary to eliminate immediate hazard(s) to study participants or when the change(s) involves only administrative or logistical aspects of the trial.

REB members involved in the research project do not participate in the review, discussion or decision.

The Western University HSREB operates in compliance with, and is constituted in accordance with, the requirements of the TriCouncil Policy Statement: Ethical Conduct for Research Involving Humans (TCPS 2); the International Conference on Harmonisation Good Clinical Practice Consolidated Guideline (ICH GCP); Part C, Division 5 of the Food and Drug Regulations; Part 4 of the Natural Health Products Regulations; Part 3 of the Medical Devices Regulations and the provisions of the Ontario Personal Health Information Protection Act (PHIPA 2004) and its applicable regulations. The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

Please do not hesitate to contact us if you have any questions.

Electronically signed by:

[redacted], HSREB Vice-Chair, 02/Sep/2022 14:25

Reason: I am approving this document

Note: This correspondence includes an electronic signature (validation and approval via an online system that is compliant with all regulations, See* Electronic System Compliance Review*)

Appendix 2. Letter of information and consent

University of Western Ontario School of Kinesiology

Letter of Information and Consent to Participate in the Research Study:

The influence of sex and body size on the validity of the Microsoft Kinect for measuring knee motion during landing

Principal investigator:

Dr. Derek Pamukoff

Student Investigator:

Jillian Neufeld

Introduction:

You are being invited to participate in a study examining the influence of sex and BMI on the validity of the Microsoft Kinect for measuring knee motion during landing. This study is initiated by Jillian Neufeld, a graduate student in the faculty of Kinesiology at Western University.

You were selected as a possible participant because you are over 18 years old, have a body mass index between 18-24.9 or above 30 and meet the screening criteria regarding physical activity and lower body injury. Should you not meet the eligibility criteria, you will not participate in any testing.

Please read this form and ask any questions you have prior to agreeing to participation.

Purpose of Study:

- The purpose of this study is twofold:
  - First, to compare knee motion during landing between females and males of different body sizes.
  - Second, to assess the agreement between the Microsoft Kinect and laboratory based 3-dimensional motion capture for measuring knee motion during landing for the sex and body size groups.
- Participants in this study are from Western University.
- We anticipate 40 participants.
- The study should take 4-6 months to complete.

Description of the Study:

If you agree to participate in this study, we ask that you attend a single session in the Biomechanics Lab in Thames Hall. The session will take approximately 45 minutes. The session will include height and weight measurements, a warm-up protocol, and a drop vertical jump task. Each of these assessments is described below.

After reading this informed consent document and asking any questions you may have about the study, you will be asked to complete a medical questionnaire. The medical questionnaire will ensure it is safe for you to participate in physical activity and that you meet the study requirements regarding previous injury. No medical information or personal history will be used in this study.

Assessment Protocols:

Height: Height will be measured using a wall mounted stadiometer and recorded in meters

Weight: Weight will be measured using an analogue scale and reported in kilograms.

Warm up: You will perform 10 body weight squats, 10 walking lunges with knee hugs, and 20 jumping jacks to prepare yourself for physical activity.

Drop Vertical Jump (DVJ): You will be outfitted with reflective markers on your lower body. These will be positioned on the tops of your feet, the back of your heels, both sides of your ankles and knees, the front of your shins and thighs, front and back of your hips and the base of your spine. With the markers in place, you will perform 3 practice DVJ trials. Each trial consists of stepping off a box, landing with both feet and immediately jumping straight up as high as possible, again landing with both feet simultaneously, then taking 3 steps forward. Following the practice trials, 5 valid trials will be filmed by the Microsoft Kinect, motion capture cameras and contact with the ground will be recorded by in floor force plates to measure knee motion during landing. In the event that a trial is not adequately captured by either the motion capture cameras or the Microsoft Kinect, the trial will be repeated.

Timeline for Testing Procedures:

Consent	20 Minutes
Height and Weight measurements	5 minutes
Drop Vertical Jump trials	20 minutes
Total Time	45 minutes

Risks associated with participation in the study:

- Lower body injury or discomfort equivalent to those normally assumed with plyometric exercise.
- If discomfort is felt at any time, you may withdraw, and no further testing will occur.

Benefits of being in the study:

- There are no direct benefits to participation in this study.
- Your participation will contribute to the assessment of the validity of a method for measuring knee motion during landing and the influence of sex and BMI on the measurement.

Voluntary Participation/Withdrawal:

- Your participation in this study is voluntary.
- You may decide to withdraw from this study by contacting the primary or student investigator at any time, including changing your mind about inclusion after

completing the study. Should you choose to withdraw, all information will be removed from analyses. Your name will remain on the master list kept by the researcher and letter of information and consent will be kept for 7 years.

- You may decline to answer any questions.
- Withdrawing from the study will not affect current or ongoing interactions with the research personnel or the University.

Dismissal from the study:

- The investigator may withdraw you from the study at any time if they believe withdrawal is in your best interest, such as in the case of discomfort or distress.
- The investigator may also withdraw you from the study if you have failed to comply with study requirements.
- Should the investigator withdraw you from the study, all of your data will be removed from the study.

Rights of the participant:

- If you are harmed as a direct result of this study, the investigators will assist you in any way they can.
- You do not waive any legal rights by signing the consent form.

Cost and Compensation:

- There is no cost associated with participation in this study.
- Each participant will receive $20 as compensation for their participation in this study.

Confidentiality:

- Participants' names and personal identifiers will not be used in any publications or reports generated throughout this study.
- Participant's height, weight, sex, and age will be collected and associated with their study identification number. The data collected is all numerical, no visual likeness will be recorded or stored. No identifiable information will be stored.
- The records of this study will be kept confidential and maintained for 7 years after the completion of the study.
- Paper records including the letter of consent and medical questionnaire will be kept in a locked file in a laboratory.
- Electronic records will be stored on an institutional hard drive and the institutional OneDrive.

- Access to the records will only be available to the study investigators, however access may be granted to Western University's Health Sciences Research Ethics Board for quality assurance purposes.

Contact and Questions:

- The primary investigator on this study is Dr. Derek Pamukoff and the student investigator is Jillian Neufeld. Any questions regarding the study may be directed to Dr. Pamukoff at [redacted] or Jillian Neufeld at [redacted].
- Questions regarding your rights as a participant in this research study may be directed to The Office of Human Research Ethics by telephone or by email.
- The Research Ethics Board oversees and ensures the ethical conduct of research studies. They are not part of the research team and everything you discuss with them will remain confidential.

The influence of sex and body size on the validity of the Microsoft Kinect for measuring knee motion during landing

Consent form

Copy of Consent Form:

You will receive a copy of this form to keep for your records and future reference.

I have read the Letter of Informed Consent for the study described above, have had the nature of the study explained to me and agree to participate. I have had all my questions answered to my satisfaction. I understand that I may withdraw from the study at any time without penalty. I agree to take part in this study.

___________________	___________________	___________________
Print name	Signature	Date (DD-MM-YYYY)

My signature means that I have explained the study to the participant named above. I have answered questions to their satisfaction.

___________________	___________________	___________________
Print name of person obtaining consent	Signature	Date (DD-MM-YYYY)

Appendix 3. Medical questionnaire

The influence of sex and body size on the validity of the Microsoft Kinect for measuring knee motion during landing

ID number:

Medical questionnaire

Please circle the appropriate response to the following questions:

Yes/No	Have you sustained a lower body injury in the last 6 months?
Yes/No	Do you have any history of lower body surgery?
Yes/No	Do you have any medical conditions that prevent you from participating in physical activities?
Yes/No	Are you recreationally active? (Participate in physical activity at least 3 times per week)

Appendix 4. Data collection sheet

The influence of sex and body size on the validity of the Microsoft Kinect for measuring knee motion during landing

Data Collection Sheet

Date: ____________________
Database ID: ____________________
Assessor ID: ____________________

Age: _________ Height (m): _________ Limb dominance: R / L
Sex: _________ Mass (kg): __________ BMI (kg/m^2): _______

Kinect measurements of knee kinematics

Knee abduction angles

	Trial 1	Trial 2	Trial 3	Trial 4	Trial 5
Angle at initial contact					
Angle at peak flexion					

Knee ankle separation ratios (KASRs)

	Trial 1	Trial 2	Trial 3	Trial 4	Trial 5
KASR at initial contact					
KASR at peak flexion					

Notes:

Version date: 31 August, 2022

Appendix 5. Supplemental analysis

Comparison of knee abduction angles at peak knee flexion with outlier removed

When the outlier was removed, the sex by BMI group by method interaction effect was not significant ($F_{1,35}$=0.035, p=0.852). However, there was a significant method by BMI group interaction ($F_{1,35}$=11.339, p=0.002). When collapsed across sex, those with high BMI had greater knee abduction than those with low BMI, but only when measured with 3D motion capture (-5.92º [-8.94, -2.90], vs. 3.20º [0.75, 5.65], p=0.005).

Table 5 Summary of descriptive statistics with outlier removed

	Low BMI Female (n=10)		**Low BMI Male (n=10)**		**High BMI Female (n=10)**		**High BMI Male (n=9)**	
	Kinect	Motion Capture	Kinect	Motion Capture	Kinect	Motion Capture	Kinect	Motion Capture
Knee Abduction PKF (º)	2.416 (3.283)	1.251 (9.241)	1.630 (5.972)	-5.167 (6.259)	.524 (5.980)	-7.695 (5.125)	5.881 (5.353)	-4.154 (3.641)

Note: PKF = peak knee flexion

Curriculum Vitae

Education

Western University

Master of Science (candidate)

Degree expected summer 2023

Faculty of Kinesiology, Integrated Bioscience, Biomechanics

Advisor – Dr. Derek Pamukoff, PhD

Thesis – The influence of sex and body size on the validity of the Microsoft Kinect for measuring knee motion during landing

University of Winnipeg

Bachelor of Science

Faculty of Kinesiology, specialization in Athletic Therapy

Sept 2014 – Jun 2018

Academic Employment

Western University

Teaching Assistant with the Department of Kinesiology. Instructing laboratory and clinical skills, lecturing and evaluating student performance for the following courses:

Biomechanics (KINESIOL 2241)

Biomechanics (KINESIOL 3353)

Athletic Injuries (KINESIOL 3336)

Clinical Employment

University of Western Ontario Varsity clinic [2022-2023]

Assessment, treatment, and rehabilitation of injuries for university athletes.

Diamond Athletic Medical Supplies Inc. [2019-2021]

Orthopedic Bracing

Fitting and customizing orthopedic braces for patients according to patient needs and prescriptions. Educating patients about their diagnoses, treatment options and braces. A high emphasis on rigid knee bracing for osteoarthritis and traumatic knee injuries

Research and Development

Piloting new orthopedic devices and providing feedback for the Research and Development team at Ossur ™, a bracing and prosthetics developer.

Administration

Insurance billing through Non-Insured Health Benefits, Manitoba Health, Manitoba Public Insurance, Workers Compensation Board of Manitoba, and other federal insurance providers

April 2019-present

Elevate Sports Therapy [2018-2021]

Owner

Clinic advertising, facility and equipment maintenance, managing booking and billing procedures and business administration

Athletic Therapist

Injury prevention, assessment and rehabilitation

<u>Athletic Therapy Field Work</u>

University of Western Ontario USports football assistant AT [2022-2023]

Football Ontario High-Performance program head AT [2022-2023]

Football Manitoba Provincial High-Performance program assistant AT [2018-2020]

Winnipeg Wild AAA U16 Hockey team head AT [2019]

Vincent Massey Collegiate football team head AT [2018-2019]

Responsibilities:

Injury Prevention

Exercise prescription and prophylactic taping or bracing

Emergency Management

Emergency Action Plan development and implementation

Coordinate care with Emergency Services

Injury Assessment

On field assessment

Removal from play and return to play decisions

Injury Rehabilitation

manual treatment

exercise prescription

Administration

Athlete medical information

Injury charting and documentation

Communication with athletes, coaches, and parents

Drug and substance education

Mentorship

Athletic Therapy Candidate Supervisor [2018-2023]

Responsibilities

Supervise Athletic Therapy Candidates during field placements with football teams

Teach emergency protocols, injury management skills, communication skills and team administrative tasks

Evaluate student performance according to the Canadian Athletic Therapy Association (CATA) core competencies

Provide feedback and scenario-based practice for Candidates to demonstrate competencies

Research Experience

Western University

Biomechanics Laboratory [2021 – 2023]

Assist with the CIHR-funded project with a long-term goal of comparing Anterior Cruciate Ligament (ACL) reconstruction techniques. Short term aims of analyzing gait patterns, landing patterns, and limb strength in ACL reconstruction patients.

Equipment

3-Dimensional Motion Analysis Camera System with Cortex Software

Visual 3D motion analysis software (C-Motion)

6 Degree of Freedom AMTI Force plates

Kinect ACL Gold software

Biodex Isokinetic Dynamometer

Supervisor – Dr. Derek Pamukoff, PhD.

Certifications

Certified Athletic Therapist (CAT(C))

Certification number 2-6024

2018 – present

Sport First Responder and Basic Life Saving

2016 – present

Memberships

Canadian Athletic Therapy Association

2016-present

Canadian MSK Rehab Research Network

2022-present

Extracurriculars and Service Activities

University of Winnipeg Kinesiology and Athletic Therapy Student Association

Faculty student council representative for Athletic Therapy program

2017-2018

Canadian Sports Center – Manitoba

Strength and Conditioning Intern [2018-2019]

Responsible for exercise instruction and program implementation for national and provincial level athletes competing in taekwondo, softball, basketball, and professional hockey.

Supervisor – Dr. Adam Decker, Ph.D.

Practical Exam Proctoring

University of Winnipeg Athletic Therapy practical exam assessment [2019-2020]

Sport First Responder (KIN3504)

Practicum Two (KIN3500)

Practicum Three (KIN4500)

Athletic Therapy program exit exam