EXTERNAL ANKLE SUPPORTS ALTER KINEMATICS AND KINETICS DURING DROP-JUMP LANDING AND FORWARD-JUMP LANDING TASKS

A Dissertation
Submitted to the Graduate Faculty
of the
North Dakota State University
of Agriculture and Applied Science

By

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In Partial Fulfillment of the Requirements
for the Degree of
DOCTOR OF PHILOSOPHY

Major Program:
Human Development and Education
Option: Wellness

March 2014

Fargo, North Dakota

North Dakota State University
Graduate School

Title

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The Supervisory Committee certifies that this ***disquisition*** complies with North Dakota State University's regulations and meets the accepted standards for the degree of

DOCTOR OF PHILOSOPHY

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ABSTRACT

This research project was designed to understand the influence of excessively restricting ankle range of motion (ROM) on knee injuries, especially non-contact anterior cruciate ligament (ACL) tear. Participating in physical activity without injuries is important to maintain physically active life style and well-being. To prevent ankle injuries, external ankle supports (EAS) are widely used in sport settings by limiting frontal plane ankle ROM; however, the EAS also restricts sagittal plane ankle ROM that could increase the risk of non-contact ACL injury by intensifying the medial knee displacement (MKD) and ground reaction force (GRF). In this research, the effects of external ankle supports (EAS) on landing mechanics were investigated among 19 physically active college-aged females. Two research manuscripts report the results of this research project.

The first manuscript investigated the effect of EAS on landing kinematics and kinetics during a drop-jump landing task. The results demonstrated the use of EAS altered the ankle displacement, total MKD, and vertical GRF; however, no relationship was observed between isokinetic plantar flexor strength and landing mechanics. The second manuscript compared the effect of EAS on landing kinematics and kinetics between drop-jump landing and forward-jump landing tasks. The result exhibited the use of EAS similarly affect ankle displacement, knee displacement, peak MKD in drop-jump landing and forward-jump landing tasks. However, the landing tasks affected the posterior GRF differently, and the EAS altered vertical GRF differently in the two landing tasks.

Overall, excessively restricted ankle ROM changed the landing kinematics and kinetics, especially MKD and GRF during landing tasks. Our findings indicate that healthcare

professionals should use EAS with care because the overly limited ankle ROM could increase the risk of non-contact knee injuries by increased MKD and GRF.

Future research should include an examination of the effect of EAS on the magnitude of ACL strain, an assessment of the strength of the other muscles, an evaluation of the muscular activation during a landing task. These studies help understand the landing techniques and strength training to reduce the risk of non-contact ACL tear among physically active population.

ACKNOWLEDGEMENT

This research project would not have been possible without the support of many people. I would like to thank the members of my dissertation committee, Dr. Bryan Christensen, Dr. Kyle Hackney, and Dr. Carlos Hawley Jr., for their knowledge and support on this project. My gratitude goes to Dr. Pamela Hansen for her invaluable assistance, support, and guidance on completing this research study. Also, thank you to NDSU's College of Human Development and Education for funding this research. My deepest love and appreciation go to my family for their understanding, encouragement, and continuous support through the duration of my academic endeavors.

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INTRODUCTION

Background

Physically active lifestyle begins at youth age and can be carried over to adulthood. Participating in sports without suffering injuries is also essential for long-term physically active lifestyle. Prevention of injury is important for sports medicine professionals. Non-contact anterior cruciate ligament (ACL) injury has been receiving more attention in recent years. Females are three to four fold more susceptible to non-contact ACL injury compared to males.[1] Factors that could lead to non-contact ACL injury is greater knee valgus angle and greater ground reaction force (GRF), and subsequent loading of the ACL during a landing or deceleration task.[2]

The strength of the plantar flexors is essential to absorb a landing force during a landing task because the foot is the first part of the body to make contact with a landing surface.[3] The range of motion (ROM) at the ankle joint and the strength of the plantar flexors are associated with GRF, and medial knee displacement (MKD) is believed to contribute to ACL injuries.[2-4] Thus, some have postulated that a smaller ankle ROM may alter the landing biomechanics,[5-7] and consequently, increase the risk of non-contact ACL injury. External ankle supports (EAS) are often used to prevent ankle injury in athletic setting. Application of EAS not only limits frontal plane movement at the ankle joint but also restricts sagittal plane ankle motions.[8] This restricted ankle sagittal plane motion could alter the landing kinematics and kinetics,[5] and as a result, may increase the risk of non-contact ACL injury.

It is important to study landing biomechanics among females along with the use of EAS. The use of EAS is common to prevent ankle injuries in athletic training settings, but the influence of EAS on the landing biomechanics has not been investigated by two-dimensional video digitizing. This study may provide a novel focus for using EAS in athletic training settings.

Framework for the Study

Physically active lifestyle in early life can help control body weight and can be carried in the adulthood. The recent increase of female sports participation[9] has also raised the number of sport-related injuries. Sports-related injuries are most common in the ankle joint followed by the knee joint.[10] Injuries to the knee joint occurred more often among male athletes due to collision-type sports, but the rupture of knee ligament including anterior cruciate ligament (ACL) was more common among female athletes, especially in basketball and soccer.[10-12] This trend was observed in both high school and in college sports.[13-15]

Although a small percentage of injuries to the knee joint required a surgical reconstruction; reconstruction of the ACL accounted for 80% of knee surgeries in general population.[16] The highest risk of ACL rupture appeared to be gender dependent. Males sustained an ACL tear most often in their 20s; on the other hand, females suffered ACL injury most often between age 11 and 20.[17] Injury rate of the ACL injury was the higher among females compared to males even though the study included contact ACL injuries commonly occurred in high-contact sports, such as football and wrestling.[18] Among college athletes in the U.S., female athletes suffered 3.25 to 4.0 times more non-contact ACL injury compared to male athlete in soccer and basketball.[1] In Europe, ACL injury was commonly occurred in soccer, (European) handball, and skiing.[18]

The risk of non-contact ACL injury was higher among females than males, teenage females are more likely to suffer non-contact ACL tear than adult females.[1,17,19,20] During the secondary growth spurt, females increased quadriceps angle (Q-angle), hip internal rotation, knee valgus angle, foot pronation, and anterior knee ligament laxity.[21-23] In addition to postural change in the growth spurt, postpubertal females demonstrated the greatest quadriceps-hamstrings ratio compared to prepubertal females, prepubertal males, and postpubertal males.[23] These postural and neuromuscular changes might lead to different kinematics and kinetics between genders during a landing task. Prepubertal males and females showed similar knee valgus angle and GRF; however, following puberty, females demonstrated greater GRF and knee valgus angle than males during a drop-off landing task.[24-26] Another gender-related factor is the fluctuation of female hormones, especially estrogen and progesterone, during the menstrual cycle.[27]

Potential ACL injury theories were ligament dominance (insufficient joint stiffness), quadriceps dominance (muscular imbalance of quadriceps and hamstrings), leg dominance (imbalance of the two lower extremities), and trunk dominance (insufficient core control to resist against the trunk lateral flexion) theories.[2] Common mechanisms of non-contact ACL injury were plant-and-cut, landing from a jump, and decelerating without change of direction with little force attenuation at the ankle joint that resulted in the internal rotation and valgus loading of the knee.[28,29]

Many previous studies have focused on kinematics of the hip and the knee joint during landing tasks.[30-35] Hence, the ankle joint that plays an important role for landing force attenuation or absorption has not been paid great attention. The foot is the first part of the body to make contact with a landing surface, and the foot position at the initial contact could decide the magnitude of sagittal plane ankle joint displacement.[5,6] Moreover, the strength of the plantar flexors is essential to absorb a landing force during a landing task.[3] The less angular displacement at the ankle joint and less strength of the plantar flexors were inversely associated with GRF.[3] Additionally, the less ankle dorsiflexion and smaller plantar flexor strength increased the MKD.[4] However, the importance of the gastrocnemius and the soleus has not been clearly explained by previous research. Smaller ankle dorsiflexion ROM and weaker gastrocnemius strength was related to a greater MKD in a descending phase of squat movement.[4] Conversely, the gastrocnemius might not be fully used to attenuate a landing force during natural landing[3]; therefore, the soleus possibly contributes to attenuate a landing force, as well. GRF not absorbed at the ankle joint would be transferred to the knee and hip joints. In addition to abnormal GRF,[36,37] one of the non-contact ACL injury mechanisms is the excessive valgus angle at the knee joint.[2] Frontal plane knee kinematics appeared to be affected by the sagittal plane ankle and knee joint kinematics.[38-40] However, the influence of the ankle joint in terms of the knee valgus and ACL loading during a landing has been neglected to date. Restricting the ankle ROM may alter the landing mechanics that is believed to increase the risk of non-contact ACL injury.

The EAS, including ankle taping and bracing, have been widely used in various sports settings. Although the main purpose of EAS is to limit the frontal plane ankle motion, ankle inversion and eversion ROM, to reduce ankle sprain, the application of EAS also restricts the sagittal plane ankle motions, plantar flexion and dorsiflexion.[8,41] This restricted sagittal plane ankle could contribute to alter the landing kinematics and kinetics[5] and could increase the risk of non-contact ACL injury.

In conclusion, physically active lifestyle begins at a young age and will be carried over to adulthood. Participating in sports without suffering injuries is also essential for a long-term physically active lifestyle. Females are more likely to sustain a non-contact ACL injury compared to males, and gender-dependent factors have been suggested. During descending phase of activities, the lower extremities must dissipate GRF so that joints are not overloaded. Therefore, ankle joint plays a significant role to attenuate landing force because the foot and ankle joint are the first body part to contact the landing surface. If the landing force is not attenuated, the rest of force will be transferred to proximal body part. However, the effects of EAS on the lower extremity landing kinematics and kinetics have not been studied to date.

Purpose of Study

Female sports participation has grown dramatically over the last forty years. With increased participation, females are more at risk of non-contact ACL rupture than males. One cause is female's landing mechanics differs from male counterparts. Smaller flexion of the lower extremity during a deceleration phase of activities and subsequent greater GRF and knee valgus might increase the risk of the ACL injury. Therefore, the first purpose of this study was to investigate whether the limited ankle ROM by application of EAS changes landing kinematics and kinetics during a drop-jump landing. Both drop-jump landing (DJL) and forward-jump landing (FJL) tasks are inevitable in many sports activities. Therefore, the second purpose of study is to investigate whether the limited ankle ROM by application of EAS and landing tasks would similarly change landing kinematics and kinetics during DJL and FJL tasks.

Hypotheses of Study

(1) Limiting ankle ROM using external ankle supports (EAS) would decrease ankle displacement, increase total MKD (tMKD), and increase vGRF during a drop-landing task.

(2) Greater isokinetic plantar flexor strength would be inversely correlated with tMKD and vGRF.EAS application will change landing kinematics and kinetics more in DJL task than FJL task.

(3) There would be difference in sagittal plane ankle displacement, sagittal plane knee displacement, peak MKD, vertical GRF (vGRF), and posterior GRF (postGRF) during landing tasks among ankle conditions controlled by EAS application.

(4) There would be difference in sagittal plane ankle displacement, sagittal plane knee displacement, peak MKD, vGRF, and postGRF between DJL and FJL tasks.

Limitations, Delimitations, and Assumptions

Limitations

(1) Menstrual cycle was not controlled even if the questionnaire included a question regarding subject's cycle of menstruation.

(2) Joint transverse-plane movement such as joint rotation was not measured due to use of two-dimension video cameras.

(3) The GRF measured was not separated into left and right lower extremity GRF.

(4) The landing tasks tested were natural landing followed by a vertical jump with a minimum instruction. This could results in relatively inconsistent results dependent on individual effort.

Delimitations

(1) Recreational physically active individuals were recruited to the study because NDSU athletics has implemented a jump-landing training among female student-athletes.

(2) Subjects were only college-aged females to avoid maturational differences among subjects.

(3) Only one type of EAS, ASO lace-up ankle brace, was used in the study.

(4) Joint torque or shear force was not analyzed.

Assumptions

(1) MKD measured in two-dimension video digitizing could represent knee valgus measured in three-dimension video analysis.

(2) Greater knee valgus increases the risk of ACL tear.

(3) Greater GRF increases the risk of ACL tear.

(4) College-aged subjects are at postpubertal stage of maturation.

Definition of Terms

<u>Abduction</u> represents a motion of a joint that is moving to the side or lateral away from the midline of the trunk in the frontal plane.[42]

<u>Adduction</u> represents a motion of a joint that is moving medially or toward the midline in the frontal plane.[42]

<u>Anterior</u> indicates a position or direction that is relating to the front or in the front part of the body.[43]

<u>Distal</u> means position or direction of the body that is situated away from the center or midline of the body, or away from the point of origin[43]

<u>Extension</u> represents a motion of a joint that is straightening or increasing the angle in a joint in the sagittal plane.[42]

<u>External rotation</u> is a rotary movement around its axis away from the midline of the body.[42]

<u>Flexion</u> represents a motion of a joint that is bending or decreasing the angle in a joint in the sagittal plane.[42]

<u>Frontal plane</u> is an anatomical plane divides the body into front (anterior) and back (posterior) sections and is perpendicular to the sagittal plane.[43]

High-tension brace (HTB) is one of the treatments in which each lace of a prophylactic lace-up ankle brace is tied with 7 – 7.5 kg tension. The tension is measured by a hand-held scale is maintained for one second during an EAS application.

Hyperextension represents a motion of a joint with excessive stretching or extension movement[44]

Inferior indicates a position or direction of the body relating to below in relation to another structure[43]

Internal rotation is a rotary movement around its axis towards the midline of the body.[42]

Lateral indicates a position or direction relating to the side, farther from the middle.[43]

Low-tension brace (LTB) is one of the treatments in which each lace of a prophylactic lace-up ankle brace is tied with 2.5 - 3 kg tension. The tension is measured by a hand-held scale is maintained for one second during an EAS application.

Medial indicates a position or direction relating to the middle or center.[43]

Posterior indicates a position or direction of the body that is relating to the behind, in back, or in the rear.[43]

Pronation is a motion of forearm and forefoot that is internally rotating movement of forearm and forefoot towards the midline of the body so that the palm is down in the anatomical position, or the plantar surface is turned outward.[42]

Proximal indicates a position or direction of the body relating to nearest to the trunk or the point of origin[43]

Rotation represents a rotational motion in which all points of a part describe circular arcs around its axis.[42]

Sagittal plane is a plane divides the body into right and left sections and is perpendicular to the frontal plane.[43]

Superior indicates a position or direction of the body relating to above in relation to another structure[43]

Supination indicates a motion of forearm and forefoot that is externally rotating away from the midline of the body so that the palm is up in the anatomical position, or the plantar surface is turned toward the midline.[42]

Transverse plane is a plane divides the body into above (superior) and below (inferior) sections and is perpendicular to both the sagittal and frontal planes.[43]

Valgus represents a position of a body part that is bent away from the midline of the body, or the outward angulation of the distal segment of a bone or joint.[44] Knee valgus may be a pure abduction motion of the tibia relate to the femur or may be combination of abduction and knee rotation.[45]

Varus represents a position of a body part that is bent towards the midline of the body, or the inward angulation of the distal segment of a bone or joint.[44] Knee varus may be a pure adduction motion of the tibia relate to the femur or may be combination of adduction and knee rotation.[45]

Acronyms

ACL: Anterior cruciate ligament

DJL: Drop-jump landing

EAS: External ankle supports

FJL: Forward-jump landing

GRF: Ground reaction force

HTB: High-tension brace

LTB: Low-tension brace

MKD: Medial knee displacement

postGRF: Posterior ground reaction force

ROM: Range of motion

tMKD: Total medial knee displacement

vGRF: vertical ground reaction force

Operational Definitions

Drop-jump landing (DJL) was completed by following steps. (1) dropping off the box (31 cm) with leaving two feet simultaneously, (2) landing with two feet on the center of the forceplate platform located 30 cm away from the box, (3) immediately performing a maximum vertical jump with raising both arms similar to rebounding task in basketball,[24] and (4) landing back to the forceplate platform. The motions of arms were not restricted during the task.

Forward-jump landing (FJL) was completed by following steps. (1) jumping off the box (31 cm) with leaving two feet simultaneously, (2) jump forward to reach the center of the forceplate platform located away at 50% of the subject's body height,[46] (3) landing with the two feet simultaneously, (4) immediately performing a maximum vertical jump with raising both arms similar to rebounding task in basketball,[24] and (5) landing back to the forceplate platform. The motions of arms were not restricted during the task.

Medial knee displacement (MKD) represented apparent frontal plane knee positions relative to the hip width. MKD was measured as a ratio to the distance between the left and right anterior superior iliac spine.[47] This valgus-like knee movement involved ankle, knee, and hip rotation that are only measured by three-dimension motion analysis. A total MKD (tMKD) was calculated by subtracting the peak MKD from initial-foot-contact MKD; that is, total tMKD represented the medial knee displacement during a landing task. Peak MKD represented the maximum medial knee movement during a landing task; that is, smaller the value of MKD demonstrated the greater medial knee movement.

LITERATURE REVIEW

Anatomy of the Lower Extremity

The lower extremity consists of foot, lower leg, patella, thigh, and pelvis. Joints including ankle, knee, and hip, are located between these body parts. Ligaments function as stabilizers at each articulation, and muscles act as movers as well as stabilizers. Each joint is dynamically stabilized by coordinated muscle functions.

Anatomy of Foot and Ankle Joint

The foot consists of seven tarsal bones (calcaneus, talus, cuboid, navicular, and three cuneiforms), five metatarsal bones, and 14 phalanges (toes). Two bones, particularly calcaneus and talus, have a significant role to transmit the body weight to the ground. The calcaneus, known as a heel bone, has a direct contact with the ground. The superior surface of the calcaneus articulates with the inferior surface of the talus, and this articulation is called subtalar joint. Most of inversion and eversion occur at the subtalar joint along with the movement at the transverse tarsal joint and plantar flexion at the ankle joint.[48] Pronation and supination of the foot occurs at the subtalar joint through anterior-posterior axis. These sideways motions of the calcaneus relative to the talus has a direct influence on the rotation of the lower leg, the tibia and fibula.[49]

The ankle joint is located between the superior part of the talus and the distal ends of the tibia and fibula. Although the axis of the ankle joint motion changes dependent of the angle of talus, the ankle joint is commonly considered as a hinge joint that allows dorsiflexion and plantar flexion of the foot in the sagittal plane. Because the posterior part of the talus is narrower than its anterior part, the posterior part of the talus loosely sits within the ankle mortise between the medial and lateral malleoli. Therefore, the ankle joint is relatively unstable when the joint is in plantar-flexed position. The ankle joint is enclosed and loosely stabilized by the fibrous capsule that is reinforced by the ligaments laterally and medially.[48,50]

The anterior and posterior muscles work as dorsiflexors and plantar flexors, respectively. Particularly, the posterior muscles or plantar flexors that collectively insert into the posterior aspect of the calcaneus are important to absorb the impact in running and landing movements. These muscles include the soleus and the gastrocnemius. Because the soleus, located deep to the gastrocnemius, crosses only the ankle joint, this is a pure ankle plantar flexor muscle.[48] The gastrocnemius has two heads that originate at the posterior aspects of the medial and lateral condyles of the femur[44]; therefore, the gastrocnemius crosses the ankle and knee joints and works as the knee flexor and the ankle plantarflexor.[48]

Anatomy of Knee Joint

The femur, tibia, fibula, and patella form the knee joint. As a result there are three articulations; the tibiofemoral joint, patellofemoral joint, and proximal tibiofibular joint. The tibiofemoral joint is formed by the distal end of the femur and the proximal end of the tibia; the patellofemoral joint is located between a femoral groove and the patella; the proximal tibiofibular joint consists of the lateral surface of the proximal tibia and the proximal part of the fibula. In this research project, the tibiofemoral joint is operationally referred as the “knee joint.”

In the knee joint, the medial and lateral condyles of the femur are articulated with the medial and lateral condyles of the proximal tibia.[48] The medial and lateral meniscus deepens the smooth surface of the tibia (tibial plateaus) so that the convex-shaped femoral condyles lie in the concave-shaped deepened menisci.[51] However, the stability of the tibiofemoral joint is still insufficient due to the configuration of the femoral condyle and shallow surface of the tibial condyles. Therefore, the stability of the knee joint is largely depending on the ligaments connecting the femur and tibia and on the surrounding musculature.[48]

Two ligament groups and two muscle groups largely contribute to the stability of the knee joint. Two ligament groups are extracapsular and intra-articular ligaments and are located between the femur and the tibia. Extracapsular and intra-articular mean outside the joint capsule and inside the joint, respectively. The extracapsular ligaments include the medial collateral ligament (MCL) and the lateral collateral ligament (LCL), and the both ligaments locate outside of the joint capsule. The MCL is a flat band-like ligament locating on the medial aspect of the knee joint and resists against knee valgus force. By contrast, the LCL is a strong cord-like ligament on the lateral aspect of the knee joint and resists against knee varus force.[48] The intra-articular ligaments consist of anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL). The cruciate ligaments are located in the center of the knee joint and cross each other obliquely within the knee joint. The ACL is located between the posterior part of the medial side of the lateral condyle of the femur and the anterior part of the tibial plateau so that it prevents anterior displacement of the tibia relative to the femur or posterior displacement of the femur on the tibia.[48]

Not only does the ACL prevent tibial anterior translation, but the hamstring muscle group also prevents the joint movement by holding the proximal tibia posteriorly. The ACL also resists against abnormal tibial internal rotation and external rotation when the knee is in a flexed position, addition to assisting MCL and LCL in resisting against valgus and varus stress.[27,52] The PCL originates from the anterior part of the lateral surface of the medial condyle of the femur and attaches into the posterior area of the tibial plateau preventing posterior displacement of the tibia on the femur or anterior displacement of the femur relate to the tibia.[27,48]

The two muscle groups are the quadriceps (the rectus femoris, the vastus medialis, the vastus intermedialis, and the vastus lateralis) and the hamstring (the biceps femoris, the semitendinosus, and the semimembranosus) groups. The quadriceps group is the main part of the anterior thigh muscles and collectively forms the patellar tendon; hence, the quadriceps is the powerful knee extensor muscles. The rectus femoris crosses two joints, the hip and knee joint, so that it flexes the hip joint and extends the knee joint. Other four vastus muscles originate from the femur and merge into the rectus femoris, so these three muscles are also the knee extensors. Posterior thigh muscles are collectively called the hamstrings. Because the hamstring muscles cross both the hip and knee joints this muscle group extends the hip joint and flexes the knee joint. Yet, the hamstring muscles are divided into two groups depending on its location; laterally biceps femoris, and medially the semimembranosus and semitendinosus. Laterally located, the biceps femoris inserts into the posterior fibular head, so it also externally rotate the tibia when the knee joint is flexed. On the other hand, medially located semimembranosus and semitendinosus insert into the proximal posterior tibia and proximal medial tibia, respectively; thus these muscles internally rotate the tibia when the knee is in flexed position.[48]

Anatomy of Hip Joint

The hip joint is a strong and stable multiaxial joint due to the ball-and-socket type structure and strong ligaments. The head of femur (ball) sits in the acetabulum (socket) of the pelvis, and the joint is enclosed by thick and strong ligaments; thus, this joint is stable but allows wide ROM.[48] More than 17 muscles are identified at the hip joint, but for the purpose of this literature of review, only several large muscles are mentioned. The hip muscles are commonly divided into three compartments; anterior, posterior, and medial compartments. Anterior compartment includes iliopsoas and rectus femoris muscles. The iliopsoas is the major hip flexor muscle, and the rectus femoris, two-joint muscle, assist the iliopsoas. The posterior compartment stores the gluteus maximus muscle, the gluteus medius, the gluteus minimus, and the external rotators in addition to the hamstring muscles mentioned above. The gluteus maximus muscle forming the main bulk of the buttocks is the largest muscle in the hip region. The main functions of the gluteus maximus are extension and external rotation of the thigh. The hip extension is also assisted by the hamstring muscles, two-joint muscles. The gluteus medius and minimus muscles almost always work together due to the same direction of the muscle fibers and the same nerve supply. The functions of these muscles are abduction and medial rotation of the thigh. The gluteus medius and minimus muscles are responsible for preventing dropping the unsupported side of the pelvis in locomotion. The external rotators collectively stabilize the head of the femur in the acetabulum and externally rotate the thigh.[44,48]

Sports Participation and Injuries to the Knee Joint

Sport participation has been growing, particularly females, in last 40 years. As female sports participation increased, the difference of injury characteristics has emerged, especially in ACL tears. This difference was most notable in soccer and basketball in the United States.

Benefits of Sports Participation in Adolescents

Participating in physical activity in adolescent has positive influence on healthy lifestyle later in life without chronic diseases, such as obesity, cardiovascular diseases, and diabetes. Although the number of high school athletes has been increasing in last few decades, the number of injuries associated with athletic participation has also been increasing and sometimes has negative influence, such as osteoarthritis in the knee joint,[53] later in life.

Sedentary lifestyle could increase the risk of chronic diseases, such as cardiovascular diseases, stroke, cancers, diabetes, and osteoporosis.[54] Conversely, physically active lifestyle decrease the risk of many chronic diseases.[55] Sports participation during childhood could predict physically active lifestyle in adulthood. This trend has been observed in both male and female.[55,56] Physical activity is beneficial to control body weight and Body Mass Index (BMI). Those who with lower BMI might have the higher risk of musculoskeletal injury compared to those who with greater BMI.[57] Therefore, continuing active lifestyle without devastating injuries is beneficial for long-term health.

Number of Female Sports Participation

Sports participation in high school has been increasing tremendously in recent years, especially among girls. The primary reason of the augmentation was probably due to the enactment of Title IX in 1972 that expanded the opportunities for girls' sports participation in education-based settings.[58] During 2011 and 2012 school year, the number of high school female athletes has increased approximately 11 fold (from 294,015 to 3,207,533) since the implementation of Title IX, and the number of athletic participation in boys has only increased by approximately 20% (from 3,666,917 to 4,484,987) during the same period.[59] The ratio between boys and girls sports participation changed from 12:1 to 1.4:1, and the girls accounted for approximately 40% of high school sports participation.[60]

The most popular sports in the U.S. for boys were football, track and field, basketball, baseball, and soccer. Basketball, volleyball, and soccer were the most popular sports among girls.[60,61] In recent years, the number participating in soccer showed the highest increase in United States among both boys and girls.[12,62] This trend has been seen not only in high school and college in the United States, but also around the world, especially among female. According to National Collegiate Athletic Association (NCAA), the number of current college men's soccer players increased by 70%; in contrast, the number of women's soccer players increased over 13 times since early 1980's.[62] The number of female soccer players has increased 54% between 2000 and 2006 worldwide.[9] These numbers, particularly in females, indicates that females may strive for physically active and healthy lifestyle.

Sports Participation and Sports-related Injury Risks

Although it has been well known that physical activity is beneficial for a healthy lifestyle, recent increase of sports participation in children, adolescents, and young adults, has also raised the risks of sports-related injuries. There are physical and physiological difference between young/ adolescent athletes and matured/ post-adolescent athletes; therefore, sports-related injury occurrence is dependent on the maturity of athletes.

Although sports-related injuries were rarely fatal, children (5 – 12 years) were the most vulnerable to sports injuries, followed by adolescents (13 – 18 years), young adults (18 – 24 years), and matured adult (> 25 years).[63] One study investigated the soccer-related injury occurrence between the range of 14 years old (under-15 group) and 18 years old (under-19 group) in Sweden. It was reported the younger-than-15 years group had the highest injury incidence (8.7/ 1000h), and the younger-than-18 years group showed the lowest incidence of injuries (4.9/1000h) and concluded that the rate of overall injuries gradually decreased as the player became older or more skilled.[64] Adolescents under 15 years old suffered more injuries (68%) with sprain and/or strain compared to children younger than 11 years old (32%) with cuts or lacerations.[65] A more recent study showed those who were younger than 14 years were more likely to injure the upper extremity, and those who were older than 15 years were likely to injure the lower extremity.[66] High school athletes suffered approximately 80% of sports injury to the lower extremities (60%) or the upper extremities (20%).[67] Among college sports, over 50% of all injuries involved the lower extremities.[68] The common injured areas in high school sports were the ankle, knee, head, back, and the common types of injuries were sprain, contusion, concussion, fracture, and muscle strain.[61] Therefore, the risk of sports related injury occurrence seemed to associate with maturation, and the adolescents are more vulnerable to sports related injuries, especially sprain and strain in the lower extremities, compared to other age groups.

Age and Gender Difference in Sports-Related Knee Injuries

Injuries to the lower extremities are common, and the knee joint was second most frequently injured body part following the ankle joint. Male athletes suffered more knee injuries than female athletes. Once it occurred, females rupture ligaments in the knee joint more than males. Injuries to the knee joints were the second most frequent (15.2%), following the ankle joints (20.9%), among high school athletes.[10] Knee injuries frequently damage one or more of the following structures; ACL, PCL, MCL, LCL, the medial meniscus, or the lateral menisci.[18] The sports-related knee injuries involving these structures accounted for more than 40% of all knee injuries.[10] The risk of sport-related knee injuries began to increase abruptly in adolescence, and this trend continued through early adulthood. Similar to the risk of injury to the lower extremities, adolescents and young athletes (10 – 24 years old) were more likely to suffer a knee injury more than matured athletes.[11,18]

Previous research showed there is no difference in knee injury occurrence between genders, and sprain and/or strain accounted for approximately 40% of all knee injuries in both genders.[11,18] There was a discrepancy, however, in the types of activity during which knee injuries occurred. Knee injuries for males happened more likely than females during sports activities (male = 62%; female =36%).[11] Among the knee injuries, the complete ligament tear involved the 8.4% for boys and 21.8% for girls.[10] It should be noted football has the highest injury rate in sport activities[69]; therefore, the gender difference in knee injury also needs to be evaluated by types of sports.

Knee Injuries in Various Sports

Sports injuries occur in various sports with different injury rates. Studies[15,70,71] found female athletes suffered more knee injuries than male counterpart in the sports that are played by both genders with the same rules. High school athletic activity injury surveillance study demonstrated the severe injury rates per 1000 athlete-exposure (AE).[71] The definition of an AE was an athlete's participation in one practice or competition, and the definition of severe injury was any injury that lead to a loss of more than 21 days of sports participation.[71] Football had the highest rate of severe injury (0.69/1000AE) in 2005 – 2007 school years, followed by boys' wrestling (0.52/1000AE), girls' basketball (0.34/1000AE), girls' soccer (0.33/1000AE), boys' soccer (0.25/1000AE), boys' basketball (0.24/1000AE), and girls' volleyball (0.15/1000AE). Of these sports, knee injuries were the most prevalent in girls' soccer (49.7%), followed by girls' basketball (44.9%), girls' volleyball (31.9%), football (25.8%), boys' soccer (23.3%), boys' basketball (20.7%), and boys' wrestling (16.7%).[71]

Soccer is the most played sport in the world.[9] According to NCAA Injury Surveillance data from 1988 to 2003, the top two soccer-related injury in competition occurred in the ankle (male = 3.19/1000AE; female = 3.01/1000AE) and the knee (male = 2.07/1000AE; female = 2.61/1000AE), and the similar trend was also observed in practice sessions.[72,73] Female soccer players were 1.06 times and 1.26 times as likely to sustain an ankle injury and knee injury, respectively, than male soccer players.[70]

The same injury trend was seen in high school basketball and collegiate basketball during the same period. The highest injured rate was at the ankle (male = 2.33/1000AE; female =

1.89/1000AE), followed by the knee (male = 0.66/1000AE; female = 1.22/1000AE).[13,14] Moreover, the male college basketball players are 0.81 times less likely to suffer ankle injury, but female players are 1.84 times more likely injured the knee joint. Among high school basketball players, the most frequent injured body part was the ankle, followed by the knee joint, and the risk of knee injury for females was significantly higher (RR = 2.29) than males.[15]

Anterior Cruciate Ligament Injury among General Population

Anterior cruciate ligament (ACL) is one of the most important ligaments in the knee joint to maintain the stability including resisting against anterior tibial translation on the femur, limiting valgus and varus stresses, and limiting hyperextension of the knee.[52] The prevalence of ACL injury was studied in general population in Sweden and New Zealand. A Swedish population-based study showed age-related trend. The risk of an ACL injury was the highest between age 21 and 30 for males, but the risk was the greatest between age 11 and 20 for females. Teenage girls had a 1.35 times higher risk of an ACL injury compared to the 20's.[17] Another population-based study completed in New Zealand reported that although the rate of surgical procedures as a result of knee injuries was only 3.9%, ACL reconstruction surgery accounted for 80%.[16] In short, an ACL injury rarely happens in general population, but adolescent, young adult males, and young adult females are at highest risk of ACL injury.

Anterior Cruciate Ligament Injury among Athletic Population

ACL injury in the athletic population is even more devastating because of its important role as a static joint stabilizer. A study in Europe reported 45% of the knee injuries involved ligamentous (LCL, MCL, ACL, or PCL) and/or meniscus tissue damage. Of those internal knee structures, ACL was the most commonly injured (50%) and occurred in soccer, ski, and (European) handball.[18] Among U.S. collegiate athletes, the highest rate of an ACL injury was reported in women's gymnastics (0.33/1000AE), followed by women's soccer (0.28/1000AE), women's basketball (0.23/1000AE), men's football (0.18/1000AE), and men's wrestling (0.11/1000AE).[68] Three women's sports reported higher ratio of ACL injuries than football and wrestling, high-contact men's sports, and the same sports played by male counterparts. Because these injury rates included an ACL injury caused by both contact and non-contact mechanisms, the result was more surprising. The non-contact injury was defined as an injury as a result of decelerating, cutting, landing, or a pivoting maneuver; in other words, no collision with another person or object was involved at the time of injury.[74]

Soccer and basketball are two sports in which an ACL injury frequently occurs. A study concluded that female soccer players are more vulnerable to ACL tear compared to male counterparts. When compared with males, females reported higher ratio of ACL injury to the total knee injury (female = 37%; male = 24%) in pediatric and adolescent soccer players.[19] The higher relative risk of ACL injury in female athletes than male athletes was also reported.[1] Among men's and women's collegiate soccer and basketball, women's soccer showed the highest ACL injury risk (0.31//1000AE) compared to women's basketball (0.27/1000AE), men's soccer (0.11/1000AE), and men's basketball (0.08/1000AE) between 1990 and 2002. The relative risk of ACL injury in females in soccer was 2.81 relates to males, and that of women's basketball compared to men's basketball was 3.38. Nevertheless, when the non-contact mechanism of ACL injury is considered, the relative risk of female ACL injury was even greater in soccer (3.25) and basketball (4.00) compared to male.[1] Another study also demonstrated college female athletes had higher relative risk of ACL injury in basketball (5.37) and soccer (9.48) compared to male athletes.[20]

Because a physically active life style of an adolescent is carried over to adulthood, it will contribute to a life with less chronic diseases. As the number of female participating in sports activity drastically increased after implementation of Title IX in 1972, the number of injury incidents, especially ACL tear, also increased. The gender difference in rate of non-contact ACL injury is significant, and studies need to investigate the reasons for the gender difference in the ACL injury.

Theories and Mechanisms of Anterior Cruciate Ligament Injury

Possible theories of non-contact ACL injury have been proposed. Although none of the theories were proven by observing non-contact ACL injury moments, each theory sounds anatomically and biomechanically understandable. Also, mechanisms of non-contact ACL injury were assessed only by two-dimension video analyses. It was found foot positions and ankle motion could be associated with altered knee joint motions, and the observation of non-contact ACL injury supported some of the possible theories.

Four Theories of Anterior Cruciate Ligament Injury

The potential ACL injury theories included ligament dominance, quadriceps dominance, leg dominance, and trunk dominance theories. The ligament dominance theory attributed to inability to absorb a landing force due to an imbalance between the neuromuscular and ligamentous control of dynamic knee joint stability.[2] This insufficient joint stability resulted in greater GRF and knee valgus and subsequently excessive loading of the ACL that functioned to restrain against anterior tibial translation and frontal plane motions in the knee joint.[2,30] The quadriceps dominance theory attributed to an imbalance between knee extensor and flexor strength, recruitment, and coordination/ co-contraction reduced joint stability during a dynamic task.[2,30] This neuromuscular imbalance, especially relatively strong knee extensors torque and weak knee flexors torque, might lead to greater knee valgus collapse and tibial external rotation.[75] The leg dominance theory was an imbalance among the two lower extremity strength, coordination, and control that could increase joint load/stress on one extremity than the other side.[2] The last theory was core dysfunction dominance theory that is an imbalance between the inertial demands of the trunk and control and coordination of the core to resist the trunk lateral flexion. This asymmetrical core control might provide more load on one lower extremity than the other.[2]

Mechanism of Anterior Cruciate Ligament Injury

Two possible mechanisms of ACL injury have been suggested; multiplanar loading with knee valgus collapse mechanism, and sagittal plane-oriented loading with anterior tibial shear mechanism.[76] Previous studies reported the common activities leading to an ACL injury were jumping, landing, planting, lateral pivoting/ twisting, or deceleration.[76-79]

It is not possible to study causes and mechanisms of ACL injury without analyzing the ACL injury. Several studies analyzed videotapes of athletes that captured the moments of ACL rupture in basketball, handball, soccer, football, cheerleading, gymnastics, and skiing.[28,29,35-37,80] Some of these studies compared the joint angles of the lower extremities and the trunk with uninjured athletes performing similar maneuvers[35,37] or analyzed the videotape with a model-based image-matching technique for better estimation of joint kinematics.[36,80] Several studies analyzed the moments when ACL tear occurred. In majority of cases (74 – 100%), players were in offensive phase, and some cases did not involve a contact to either the upper or lower body prior to ACL tear (40 - 72%).[28,36,81] Three major tasks performing during ACL tear were performing cutting maneuver accounted for 13 – 70%, performing one-leg landing was involved in 20 to 30% of cases, and performing two-leg landing was involved in 30% of cases.[28,36,81]

Approximately 80% of women and 20% of men were in a deceleration phase.[42] In addition, 19 out of 20 cases were considered as out of balance or perturbation during ACL injury.[28] Hence, it may be speculated that single-leg planting to change a direction during a deceleration phase might be contributing to an ACL tear. In some studies,[28,35,37] foot position at initial contact on the ground, sagittal plane joint angles of the foot, ankle, knee, hip, and trunk (viewed from the side), and frontal plane joint angles of the knee, hip, and trunk (viewed from the front) were evaluated.

Visual observation found that the foot was planted lateral to the knee (the imaginary line from the center of the knee perpendicular to the ground), the knee at initial contact was slightly flexed and in valgus position, and the tibia was either internally or externally rotated.[28] A case-control study found that ACL injured subjects landed with the hindfoot (heel-landing) or with the entire foot (flat-landing).[37] Moreover, ACL injured subjects demonstrated less plantar flexion of the ankle, more flexed hip joint, and less trunk flexion at initial contact phase of landing compared to control subjects. The lower extremity kinematics were not different between the groups only in the sagittal plane knee angle.[35,37] Results were not consistent in the knee flexion angle at the initial contact between a visual observational study and two-dimension video analyzing studies. Although knee and hip frontal plane position were not different between the groups, the lateral flexion trunk angle relative to the vertical line was greater in ACL-injured group. The uncontrolled lateral trunk flexion might contribute to the knee valgus during an ACL injury.[35,37] The major limitations of the two-dimension video analysis were that the exact moment of an ACL rupture could not be determined and that the accurate joint rotation angles were not measured in the transverse plane. Still, the studies concluded that the video analysis revealed the commonly observed mechanism of noncontact ACL injury was “valgus collapse” of the knee joint. The valgus collapse is a combination of knee abduction, external rotation of the tibia, and internal rotation of the femur.[28,35,37,81] Two phases commonly evaluated were a phase at initial contact of the foot with the ground and a phase at immediately after the initial contact. Researchers found that a rapid knee valgus loading occurred within 40 to 50 milliseconds after the initial contact; thus, knee collapse could be a predominant factor of a non-contact ACL tear.[35,36,81]

Although downhill skiing is not commonly reported in ACL injury studies in U.S., it is popular in Europe and often reported in ACL injury research. Common mechanisms of an ACL injury among world-class alpine skiers were studied by analyzing 20 video-recorded ACL injuries. At the moment of an ACL tear, 60% of injury occurred when the skier was attempting to change direction, and 85% of cases the boot-ski binding were not released so that no body weight was sustained and absorbed at the ankle joint.[29] The study suggested three common mechanisms of ACL injury in downhill ski; slip-catch mechanism, dynamic snowplow, and landing back weighted. In slip-catch mechanism, a nearly straight knee of the outer ski was abruptly forced into flexion, internal rotation, and valgus. In dynamic snowplow, a skier was out of balance with more weight on only one leg; subsequently, the loaded knee was forced into internal rotation and/or knee valgus.[29] The slip-catch and dynamic snow-plow mechanisms were similar to the multi-plane loading with knee valgus collapse mechanism and frequently involved internal rotation and valgus loading of the knee. These mechanisms are common mechanisms of an ACL injury during planting-and-cutting, landing, and decelerating. In the landing back weighted mechanism, the skier landed with little flexed knees and on the tails of the ski.[29] This mechanism is corresponding to the decelerating and landing without changing direction observed in other sports. Although it is difficult to investigate an actual ACL injury in video analysis or in a prospective study because the number of subject is always limited due to low injury rate as described above, analyzing injuries based on actual incidents provide significant impact on future studies. Nevertheless, relatively low frame rate (generally 30 or 60Hz) in standard video recordings used in video analyses were not clear enough to capture accurate images of rapid athletic movements.[82]

Potential Factors for Non-contact Anterior Cruciate Ligament Injury

The following factors contribute to non-contact ACL injury have been suggested; age and maturation (prepubertal vs. postpubertal), neuromuscular characteristics, gender (postpubertal males vs. postpubertal females), anatomical factors (femoral intercondylar notch width, tibial slope etc.), fatigue effect, hormonal levels, and leg dominance.

Maturation and Gender Difference in Postural Characteristics

Females were more likely to suffer a non-contact ACL injury than males, and the non-contact ACL injury rate was higher among teenage girls compared to adult females.[1,17,19,20] It should be speculated that there might be a gender-related and maturation-related factors regarding non-contact ACL injuries. A case study[83] followed an adolescent girl and reported that the female ALC injured athlete was in postpubertal stage at the time of injury. Prior to the injury the female athlete increased in height and body mass index (BMI) but decreased in hip abduction angle and hamstring muscle strength. In addition, she presented limited change in quadriceps muscle strength. Hence, the subject relatively decreased the knee flexor-to-extensor ratio and elevated center of mass.[83] The result of this case study was further investigated by several studies[21-24] that examined the alteration of kinetics and kinematics between different stages of maturation.

There are obvious physique changes with maturation, especially during the second growth spurt or puberty. A cross-sectional study[21] compared postural and anatomical differences in males and females between different maturation stages instead of chronological ages in adolescent. The researchers assigned the subjects into three categories based on self-administered Tanner's 5-stage maturation stage classification; prepubertal (stage 1) and early pubertal (stage 2) group combined, mid-pubertal (stage 3) and late pubertal (stage 4) group combined, and postpubertal (stage 5) group.[21] Some variables changed with maturation were gender dependent; in other words, anatomical and postural characteristics changed differently in males and females during growth spurt, and the difference might eventually influence on the different non-contact ACL injury rate. Genu recurvatum (hyperextension of the knee joint), quadriceps angle (Q-angle), anterior knee laxity, and foot pronation reduced with maturation stages in both genders; however, the amount of decrease in Q-angle and anterior knee laxity were greater in males during the same growth period. The pelvic anterior tilt and tibial torsion, by contrast, increased in both genders during the same period. In postpubertal group, female demonstrated greater Q-angle, anterior knee laxity, hip internal rotation, and tibiofemoral (knee valgus) angle compared to males.[21] Although the researchers did not measure the width of pelvis, another study[22] reported that female had wider pelvic width than male, which subsequently increases hip varus along with femoral anteversion, increased knee valgus and Q-angle, and foot pronation following the growth spurt.

In conclusion, postural characteristics of post-pubertal female include greater anterior knee laxity, greater knee valgus, and external rotation of the tibia associated with hip internal rotation and foot pronation. These characteristics are similar to the mechanisms of non-contact ACL injury.

Maturation and Gender Difference in Muscle Strength and Ligament Laxity

In addition to apparent physique changes, neuromuscular characteristics and ligamentous laxity also change during puberty. In the prepubertal stage, no gender difference in quadriceps and hamstrings strength was observed. During puberty, both boys and girls increased the quadriceps and hamstring isometric strength, but the rate of increased muscle strength were greater in boys (179% in quadriceps, 148% in hamstrings) than girls (27% in quadriceps, 44% in hamstrings). As a result, matured girls demonstrated greater quadriceps-hamstrings (Q-H) ratio (2.06) compared to prepubertal girls (1.73), prepubertal boys (1.58), and matured boys (1.48).[23] Also, matured boys demonstrated increased both quadriceps and hamstrings peak torques during the puberty, especially in late pubertal stage. Girls, however, did not show significant changes in quadriceps or hamstrings peak torque.[24]

Moreover, boys and girls alter ACL laxity differently during the second growth spurt. Although there was no difference in knee joint anterior laxity between genders prior to the puberty, the knee anterior laxity dramatically decreased in boys but increased in girls.[23,83] Therefore, while boys gained muscular strength and ligament stiffness during puberty, girls increased Q-H ratio and ACL laxity during the same period. These postural, muscular, and ligamentous characteristics during puberty could lead to gender differences in kinematics and kinetics during physical activities.

Difference with Maturation in Kinetics and Kinematics during Activities

Gender differences were also observed in kinetics and kinematics during landing and jumping activities during puberty. Although it is not conclusive, it has been believed the higher GRF corresponded to greater ACL loading.[80] In the prepubertal stage, boys and girls had similar quadriceps and hamstring activities and GRF in a vertical jump task targeted to their 50%-

maximum vertical jump height.[84] In the post-pubertal stage, matured subjects showed greater hamstring muscle activity at pre-landing phase and longer time to reach the peak GRF compared to prepubertal subject.[84,85] Hence, gender did not affect the muscular activities and GRF in the 50%-maximum vertical jump activity.

Meanwhile, gender seemed to influence landing pattern during a drop vertical jump task. Postpubertal boys increased vertical jump height and decreased in landing GRF, whereas girls did not change GRF with maturation during a 31-cm drop vertical jump task.[25] Although both genders lowered GRF loading rate, postpubertal boys displayed smaller landing GRF compared to postpubertal girls.[25] This trend in GRF was also observed in another study, though GRF loading rate showed no difference between pre- and post-pubertal stages.[84] As a result, while boys increased the ability to attenuate landing force, girls did not change force-attenuating ability with maturation.

In addition to the physique development and the following kinetics alteration, kinematics during drop vertical jump and vertical jump tasks changed during the second growth spurt. Before puberty no gender effect in the maximum knee valgus angle was observed.[24,84] Following the pubertal period, boys reduced the maximum knee valgus during landing, but girls did not change or even increase the knee abduction angle.[24,26,84] As the result, postpubertal boys exhibited less knee valgus angle than girls; in other words, girls demonstrated more knee valgus movement than boys with maturation during a drop vertical jump task or 50%-maximum vertical jump task. However, similar to kinetics no gender difference was observed before pubertal, and the knee valgus angle was reduced with maturation in both genders in 50%-maximum vertical jump task.[84] Therefore, 50%-maximum vertical jump task might have not been enough to observe kinetic and kinematic difference between genders.

It should be noted that some studies used two maturation stages (pre- and post-puberty);[25,26,84,86] others classified the stages into three stages (pre-, early, and late/post-puberty)[24], or four (pre-menarchial girls, after menarchial girls, boys younger than 13 years old, and boys older than 14 years old).[23] In those studies, several limitations were reported. Other potential factors of an ACL injury, such as hormone levels, femoral notch width, Q-angle, or foot pronation were not controlled, thus, maturation and gender influence might explain the disparity in ACL injury rate between genders.

Fatigue and Knee Joint Kinetics and Kinematics

Even though the athletes are trained and conditioned to minimize fatigue during a match, one epidemiological study reported the highest rate of knee injuries occurred in the final 20 minutes of a competition (number of knee injury incidents; 0-20min = 17, 21-40min = 25, 41-60min = 26, and 61-80min = 32) in rugby football.[87] This result might indicate that sport-related injury may be associated with fatigue that would alter the kinetics and kinematics of the lower extremity during landing tasks. Because studies that investigated pre- and post-fatigue conditions used different neuromuscular fatigue protocols, the result demonstrated various kinetic and kinematic alterations.[88-91]

Following an isokinetic fatigue protocol, both males and females demonstrated delayed contraction time in the hamstrings and calf muscles, but only females showed delayed onset of the vastus medialis activation without changing ACL laxity.[92] Interestingly, the GRF exhibited wide range of alteration following a fatigue exercise; decreased GRF,[88,89] no difference,[90] and increased GRF.[91] One study that observed a decreased GRF reported no difference in maximum loading rate during a single-leg landing task possibly due to other factors besides fatigue.[89]

To attenuate or absorb GRF more or less effectively, the joints kinematics of the lower extremity need to be altered. As for the hip joint, fatigue protocols did not affect the angle of the hip flexion at the initial contact, but the maximum flexion angle increased[90] or did not change.[89] In the frontal plane, the hip adduction-abduction angles showed no fatigue effect.[90,93] As for the knee joint, the fatigue effects on the knee joint kinematic changes were also controversial. The results revealed knee joint flexion angle increased,[89,90] no difference,[88,91] or even decreased[94] after fatigue. However, the maximum valgus angle and tibial rotation after fatigue showed no fatigue effect or increase following fatigue.[90,91,93] Interestingly, the ankle joint kinematic change after fatigue was consistent among researches; maximum dorsiflexion angle and foot pronation intensified due to fatigue of the lower extremity muscles.[89-91,93]

Therefore, fatigue might influence the lower extremity joint kinematics more in the frontal plane knee joint movement and the sagittal and frontal plane ankle joint movement than kinematics of the hip joint in both frontal and sagittal plane and the knee joint on the sagittal plane. This fatigued neuromuscular function and altered joint kinematics during landing tasks could be fatigue-protocol or landing-task dependent. Still, the altered joint kinematics and the unchanged GRF might indicate that the subjects possibly adjusted their landing techniques at the ankle and the knee joint to effectively attenuate landing load. Another interesting explanation of increased ankle dorsiflexion that should be mentioned was not associated with fatigue. The gastrocnemius lost the plantar-flexing moment because it is a two-joint muscle crossing the ankle joint and the knee joint; therefore, as the knee flexion increased, plantar flexion might rely on the soleus muscle strength.[89]

Anatomical Factors in the Knee Joint

Intercondylar notch width, Q-angle, and tibial slope have commonly been studied by case-control-study to investigate any of those anatomical structure differences, but the influence of intercondylar notch on ACL injury has been inconclusive.[95-98] The width of femoral intercondylar notch was generally narrower in females compared to males regardless the height of the subjects. The researchers concluded that the relationship between the ACL injury rate and the width of intercondylar notch disappeared when the size or diameter of ACL were taken into account. The result lead to a conclusion that the ACL injury rate might attribute to factors other than the width of intercondylar notch.[95] This result was supported by a recent study that compared ACL-injured group to no-injured group among females. The study revealed that there were no difference in standing Q-angle, the width of the pelvis, or the width of the intercondylar notch between injured and non-injured groups.[96]

Instead of the intercondylar notch width, other case-control studies[97,98] measured other anatomical variables including tibial plateau slope and depth, in the knee joint to compare the case-control and gender differences. In both genders, ACL-injured subjects had more posterior-directed tibial plateau slope and shallower medial tibial plateau depth for the femoral condyle compared to the control group.[97,98] Among control groups the female demonstrated more posterior-directed tibial plateau slope and shallower medial tibial plateau depth.[97] Therefore, the posterior tibial slope and shallow tibial plateau depth might be more important than the intercondylar notch width in different ACL injury rate between males and females.

Leg Dominance and Anterior Cruciate Ligament Injury

Researches revealed kinetic and kinematic asymmetries were associated with leg dominance during landing tasks.[82,99] Previous studies investigated leg dominance as a possible factor in non-contact ACL injury, but the non-contact ACL injury occurrence regarding the side of leg dominance was still debatable. Some researchers found non-contact ACL injuries occurred more on the non-dominant leg,[100,101] other researchers concluded the leg dominance could not predict the side of non-contact ACL injuries.[74,99,102]

In ACL injured athletes, males were more likely to tear the ACL on the dominant leg (26 - 55%); by contrast, female athletes were likely to suffer ACL rupture on the non-dominant leg (63 - 68%).[100,101] Therefore, non-contact ACL injuries in female might be associated with the leg-dominance, and this difference was possibly due to decreased in strength, activation, or proprioception in the non-dominant leg.[101] On the contrary, several other studies concluded that there was no association of leg dominance in non-contact ACL injury. Researchers found no association between side of injury and the dominant leg and no difference in genders between the injured and the dominant legs.[74,99,102] Therefore, the side of non-contact ACL injury could not be predicted by merely determining the dominant leg by only asking which leg was the kicking leg.

Hormonal Fluctuation and Anterior Cruciate Ligament Injury

Although the physique and neuromuscular development that occur in both genders during puberty are often discussed in literatures, only the female population experiences menarche. The menarche usually occurs at the onset of the second growth spurt, and this hormonal alteration might influence on gender difference in the non-contact ACL injury rate. During a menstrual cycle, the levels of ovarian hormones fluctuate, particularly estrogen and progesterone.

Menstrual cycle is divided into three phases; the menstrual phase, the follicular phase, and the luteal phase. Although duration in each phase varies individually, it has been reported the average duration of one menstrual cycle is 28 days (ranged from 24 to 35 days).[27] During the menstrual period that lasts roughly five days, a sudden reduction of ovarian hormones occurs.[27]

The second phase, the follicular phase, begins following the menstrual phase and lasts from day 6 to 13. During the follicular phase, follicular development stimulates estrogen secretion; consequently, the estrogen level reaches the highest.[27] At the end of the follicular phase, the sudden surge of the luteinizing hormone concentration occurs. As the peaky luteinizing hormone increase subsides, ovulation occurs and lasts for 24 hours during which the estrogen level significantly decreases.[103] Following the ovulation, the luteal phase begins and lasts from day 15 to 28. During the luteal phase, the levels of estrogen, progesterone, and relaxin increase.[27] If implantation does not occur, the levels of these hormones return to the lowest level, and the menstrual phase begins.[27,103]

The relationship between the fluctuation of ovulatory hormones and non-contact ACL injury has been investigated, but it has been difficult to conclude a cause-and-effect relationship. Among females who were not on oral contraceptives, non-contact ACL injuries were recorded more in the follicular phase (2.5 to 3 times) than the luteal phase, especially five days before the ovulation occurs. In female who were taking oral contraceptives, the similar relationship was observed but only a trend.[103-106] The non-contact ACL injury rate was lower in the oral contraceptive users; physically active females, however, might not have normal hormonal profile merely because they have normal menstrual cycle.[106] These results did not find the cause-and-effect relationship between the menstrual cycle and the increased non-contact ACL injury.

The ACL laxity was studied by measuring the amount of anterior tibial translation and the levels estrogen and progesterone. One study[27] reported that ACL laxity was the greatest in the luteal phase followed by the follicular phase and the menstrual phase. In this study the researchers divided the menstrual cycle into three phases; the menstrual, follicular, and luteal phases. Another study[107] used four phases; menses, follicular, ovulation, and luteal phases. The result was that although the difference was not significant, compared to the mid-follicular and mid-luteal phases, the ACL laxity was higher by 10% at the ovulation phase. In addition to the laxity of ACL, by calculating from a GRF and vertical acceleration of the center of mass following a single-leg hop, musculotendinous stiffness of the lower extremity decreased by 4.5% at the ovulation phase compared to the beginning of the follicular phase.[107] Still, one major limitation of ACL laxity studies was that the subjects did not suffer ACL rupture during the investigation. Hence, although the results were noteworthy, it should not be concluded that the greater ACL laxity and lower musculotendinous stiffness increase the risk of non-contact ACL injury.

Lower Extremity Kinematics and Kinetics during Landing

During a landing, closed-kinetic-chain mechanism of the body functions as a shock absorber. Joints in closed-kinetic-chain system are affected by motions at other joints that are controlled by muscles. The foot is the first part of the body to contact on a ground during a landing task, and the impact that is not absorbed is transferred from the distal joint to the proximal joint through the ankle, knee, and hip joints.[108] In other words, the work at the distal joint affects the load at the proximal joints. After all, joints and muscles of the lower extremity need to be resistive against GRF to successfully attenuate or absorb the landing force without collapsing joints and losing balance. Greater stress at a proximal joint is believed to be a result of a poor shock absorption ability of the distal joint and muscles.

Ankle Joint and Landing Biomechanics

Pronation of the foot is essential to dissipate impact from the landing surface. Although the results appeared to be landing-task dependent, foot pronation might contribute to absorb landing force. Frontal plane motion of the ankle, such as supination and pronation, might play a significant role to attenuate the landing force[82] because the ankle is the first joint to transfer the impact to the proximal joints during landing tasks. The foot becomes more flexible by pronating at the subtalar joint and unlocking the mid-tarsal joint to absorb the landing impact from the ground.[109] Despite the logical concept of frontal plane ankle motion, the influence of foot pronation and supination has not been conclusive. No difference in magnitude of GRF or rate of loading was observed among pronated-foot (GRF = 3.44 BW; loading rate = 0.05 BW/ms), supinated-foot (GRF = 3.57 BW; loading rate = 0.06 BW/ms), and neutral-foot (GRF = 3.65 BW; loading rate = 0.06 BW/ms) subjects in a drop landing task.[110]

Another study, however, reported females showed greater vertical and posterior GRF than males possibly due to different ankle and knee kinematics during a drop landing task.[38] At the initial contact phase of a drop landing, no difference were found in the lower extremity joint kinematics between genders, including the ankle plantar flexion and foot pronation. On the contrary, the peak joint angles were significantly different between genders. Females demonstrated greater peak ankle dorsiflexion (females = 32.7°; males = 23.8°), foot pronation (females = 20.9°; males = 1° supination), and knee valgus (females = 24.9°; males = 1° varus) angles than male subjects. As the results, females exhibited larger vertical GRF (females = 4.71 BW; males = 3.51 BW) and posterior GRF (females = 0.78 BW; males = 0.19 BW) than males during the drop landing task.[38] The different results between two studies might have attributed to the study methods. When the landing task was conducted by landing from 30cm with bare feet, no difference was observed between pre-measured foot conditions.[110] The other study was conducted by landing from 60cm while wearing running shoes, and the foot movement (pronation and supination) was measured during the landing task.[38] It needs to be noted that the importance of the pronation was not isolated from ankle dorsiflexion and knee valgus; therefore, the clear influence of foot pronation was still controversial.

Muscular weakness and functional stability in the ankle joint altered landing biomechanics, possibly to compensate the lack of ankle function. The muscular weakness might reduce the time to reach peak force. Functional instability of the ankle joint following an ankle sprain was associated with weaker ankle muscle strength, pain in the joint, and less ligamentous stability.[111] Subjects with ankle functional instability were compared to subjects with no ankle instability, and the result was that peak vertical GRF was not different between functional instability (GRF = 5.01BW) and control (GRF = 4.77BW) groups. Although vertical GRF increased more quickly 30 millisecond after initial contact in the instability group, and the time to reach peak GRF was also not different between functional instability (40.0 ms) and control (45.8 ms) groups.[111]

A possible reason of these vertical GRF difference was that the subjects with ankle instability could not overcome the initial impact due to muscle weakness or delayed muscular activation. In other words, the instability subjects might have attempted to absorb the landing impact, but the ankle musculature was not functionally used to attenuate the initial landing force. Another possible reason was that instability group subject might have landed differently than control group due to instability and pain. Thus, the strength of the ankle joint and muscles might play important role in landing tasks.

In summary, frontal plane foot kinematics did not influence the GRF. In 30-cm drop landing did not affect ankle kinematics. In 60-cm drop jump, however, female subjects demonstrated greater peak joint angles in foot pronation, dorsiflexion, and knee valgus in addition to greater GRF. Also, weak plantar flexor muscle could not attenuate landing force 30 ms after initial foot contact. These results indicated not only ankle displacement but also plantar flexor strength was essential to attenuate landing force.

Ankle Plantar Flexors and Ground Reaction Force

The function of ankle plantar flexors during landing tasks was investigated by several researchers. The GRF must be overcome or absorbed by muscles of the lower extremities; ankle plantar flexors, knee extensors, and/or hip extensors.[6,37,108] The ankle plantar flexors were particularly important to absorb GRF during landing tasks.[3-6] When comparing greater knee-flexion landing (soft landing) to less knee-flexion landings (stiff landing), the ground impact needed to be absorbed more by the joint of the lower extremity.[3,5,6]

Two types of drop landing technique, soft landing and stiff landing, from 59cm high platform were compared. In this study,[5] the landing techniques were divided only by the knee flexion angle during the landing task. Soft landing was a landing with greater than 90° peak knee flexion angle (mean angle = 117°); stiff landing involved less than 90° knee flexion (mean angle = 77°) during the landing task.[5] At the foot contact, the joints of lower extremity were prepared to absorb the landing impact. In the soft landing technique, the hip and the knee joints were more flexed than the stiff landing as expected, but the ankle joint was less plantar flexed at the foot contact. As a result, soft landing presented lower vertical GRF (= 2.25BW) than stiff landing (GRF = 3.16BW), possibly due to effective force absorption. The percentage of absorbed landing force by the hip and knee joints in soft landing were approximately 5.8% and 6.2%, respectively, greater than stiff landing. In contrast, the ankle joint performed less percentage to absorb the landing force in soft landing by 13% compared with stiff landing.[5] This indicated that in soft landing, quadriceps muscles, hamstring muscles, and gluteus maximus performed more to absorb kinetic energy. The plantar flexor muscles might be more activated in when the knee flexion was less involved during a landing. The importance of plantar flexor muscles was also reported in another study.[6]

During a landing from 30cm-high platform, male subjects displaced the hip joint, knee joint, and ankle joint by 4.0°, 12.9°, and 26.7° in sagittal plane, respectively.[6] As a result, the percent of work at each joint to absorb the landing energy was 16.2%, 5.7%, and 78.2% at the hip, knee, and ankle joint, respectively. Comparing male subjects, female subjects demonstrated less joint displacement at the hip (1.6°) and knee (8.3°), but the ankle joint displacement (27.1°) was similar to male's value. Although the relative energy absorbed at the hip joint (9.7%) and the knee joint (5.7%) were not different from the males, the ankle joint energy absorption rate was greater in female (88.3%). Still, the normalized maximum vertical GRF in female was 9% greater than male.[6] These results indicated that in the landing with less hip and knee flexion (stiff-type landing), the work at the ankle joint might increase to accommodate the total energy absorption.

The importance of plantar flexor muscles in stiff landing technique was also reported by another landing study.[3] Four types of drop vertical landing from an 30.5cm overhead drop bar were compared.[3] The key result was that as the knee stiffness increased, the plantar flexor muscles became more important to attenuate the landing force. Natural landing showed greater knee flexion than three other stiff-knee landings; however, stiff-knee landing with conscious plantar flexors activation demonstrated 10° more plantar flexion and recorded the least vertical GRF. Stiff-knee plantar flexion landing also showed the greatest Achilles tendon force during the vertical drop landing task. In contrast, stiff-knee heel-first landing showed the lowest Achilles tendon force and the highest vertical GRF.[3] The author believed that the gastrocnemius was not fully utilized in normal landing technique because stiff-knee landing with plantar flexors activation resulted in lower vertical GRF than natural landing technique. Consequently, the gastrocnemius, a two-joint muscle, might become slack when the knee was flexed.[3] The gastrocnemius was loosened when the knee joint was flexed, but the soleus, a single-joint muscle, may work as a primary plantar flexor. This conclusion could lead to that a future research should investigate the soleus strength and activation that may get more involved to attenuate landing impact in soft-type landing tasks.

A landing with greater knee and hip flexion angle showed lesser GRF than a landing with lesser knee and hip angle. Still, ankle plantar flexors played a significant role in attenuating landing force to compensate smaller proximal joint flexion angles. This was supported by another study that reported that ankle plantar flexion was more involved when the knee and hip flexion was limited. Because gastrocnemius is a two-joint muscle, in theory, as knee joint flexed, the muscle tension of gastrocnemius is lost, and the soleus need to produce counter force against landing force.

Ankle Range of Motion and Ground Reaction Force

In addition to the plantar flexors, the ankle displacement, particularly dorsiflexion, during a landing task were important to provide enough time for plantar flexors to generate a counter force. During a toe-landing task from 40 cm height, the GRF forced the ankle joint into dorsiflexion, and the eccentrically contracting plantar flexor muscles counteracted the dorsiflexion to absorb GRF. Thus, sagittal plane ankle displacement and plantar flexors played important roles during a landing task.[7,108] When a toe-landing was compared to a heel-landing, the ankle dorsiflexion was more than eight times greater in toe-landing technique (44°) than heel-landing technique (5°). The peak GRF was approximately 60% greater in heel-landing (GRF = 10.1BW) than in toe-landing (GRF = 6.2BW) probably due to difference in the ankle position at the initial contact in heel-landing (30.1° dorsiflexion) compared to the toe-landing (10.8° plantar flexion).[7] The non-contact ACL injury was analyzed by reviewing recorded injury moments. The result revealed that all ACL injured subject had heel landings or flat-foot landings; in contrast, the control subjects landed with the forefoot or a combination of the forefoot and the midfoot.[37] These results could indicate that heel-landing technique might be associated with the greater risk of non-contact ACL injury due to lack of ability to attenuate landing impact force by plantar flexors.

Another study also found that greater ankle displacement decreased vertical GRF during a landing task from 30.5 cm height.[3] A landing with a greater ankle plantar flexion prior to the foot contact (GRF = 4.1BW) exhibited approximately 40% smaller vertical GRF compared to less ankle plantar flexion at the initial foot contact (GRF = 6.7BW). It is interesting that vertical GRF was not different between natural landing with knee flexion (GRF = 4.3BW) and stiff-knee landing with plantar flexion (GRF = 4.1BW).[3] The result indicated that greater plantar flexion and subsequent greater ankle displacement maximally utilized the plantar flexor muscle to absorb vertical landing force.

The flexibility of the plantar flexors seemed to affect the loading stress on the knee joint during a descent phase of single-leg squat.[112] Comparing subjects with patellar tendinopathy and subjects with no patellar abnormalities, the group of patellar tendinopathy was believed to be exposed to greater knee loading stress. The patellar tendinopathy group had smaller ankle dorsiflexion ROM measured by a weight-bearing lunge test compared to the control group. The author concluded the weight-bearing dorsiflexion ROM with greater or smaller than 45° appeared to differentiate normal patellar tendon and pathological patellar tendon.[112] In other words, the ankle with less dorsiflexion could not absorb the GRF effectively, and the force was transferred to the proximal joints.

Greater plantar flexion at initial foot contact allowed the ankle joint more time to absorb the landing force accompanied by plantar flexor activation. In other words, the ankle displacement might affect the ability to force attenuation. This logic was supported by a study of tendinopathy that found greater plantar flexion flexibility could transfer smaller force to the knee joint.[112] Therefore, ankle ROM in sagittal plane may be important to absorb landing force.

Roles of Calf Muscles during Landing Tasks

Although the previous researchers concluded the plantar flexor muscles played an important role to absorb landing force during landing tasks, the roles of gastrocnemius and soleus muscles have been debatable.[3,113] One study[3] concluded that the gastrocnemius might not function in a landing with knee flexion because the muscle became flaccid when the knee is flexed. Therefore, the soleus muscle might be more important than the gastrocnemius during a landing. Another study[113] found an inverse association between knee-extended passive dorsiflexion ROM and GRF. The greater knee-extended passive ROM was also correlated with knee and hip displacement, which indicated the GRF was attenuated by greater sagittal plane joint displacement. However, the knee-flexed passive dorsiflexion ROM had no association with GRF or other lower extremity kinematics during the landing. These results lead to a conclusion that greater knee-extended passive dorsiflexion ROM and greater joint displacement at the knee joint might be associated with greater force absorption ability of the gastrocnemius.[113] Future research should compare the role of the gastrocnemius and the soleus during a landing task.

Regarding calf muscle activity, no gender difference was observed in the gastrocnemius, but soleus activity was higher among females.[114] Male seemed to absorb loading force in all lower extremity joints, whereas females tended to absorb the force by primarily knee and ankle joints. In addition, females demonstrated greater force loading rates and greater peak angular velocities compared to males.[6,115,116] These different landing techniques were related to that females could less effectively attenuate loading force than males by 24%.[6] Muscular activity and generated force were correlated with muscular stiffness that provide dynamic stability in the joints.[117] Hence, neuromuscular control deficit and less joint stiffness in the lower extremity might be related to inability to produce sufficient torque on the joints and might result in ligament-dominance landing.[117,118]

Even though subjects with excessive medial knee displacement demonstrated greater muscular strength in hip extensors and hip external rotators, they showed lower plantar flexor strength during a squat movement.[4] The excessive medial knee displacement group had approximately 17% decreased plantar flexor strength compared to the control group. The authors explained the link between decreased plantar flexor strength and greater medial knee displacement. The medial head of the gastrocnemius could not be strong enough to resist against tibial external rotation.[4] As a result of tibial external rotation, the greater medial knee displacement was recorded. Still, this result was investigated during a squat movement that was much slower movement than landing task, GRF was not measured, and the muscular strength was measured in isometric strength.

Although the ankle plantar flexors counteract the landing force, the importance of two muscles in the plantar flexors was still inconclusive. Nonetheless, the weakness of plantar flexors could not provide sufficient ankle joint stiffness to resist against landing force, and the residual energy would be transferred to the knee joint. If the knee joint is unable to provide sufficient joint stiffness, the energy could be attenuated by the ligaments in the knee joint. In addition, the weakness of the plantar flexors was related to excessive MKD in squat movement that was similar to landing task. Therefore, both joint displacement and associated muscle strength are important to effectively attenuate landing force.

Duration of Landing Phase and Ground Reaction Force

Not only the maximum joint flexion angle during a landing task but also the time of landing phase was associated with the GRF. Studies revealed that shorter the time between the initial foot contact and the maximum joint flexion might be related to the greater GRF.[5,6] Durations of landing phase was measured in two parameters; one is the time between the initial contact and peak flexion,[6] the other is the time to reach peak GRF.[5] Both studies revealed similar results.

The duration to peak GRF was shorter in stiff landing (85 ms) compared to soft landing (126 ms)[6], and the duration to peak flexion of the ankle, knee, and hip joints were also shorter in female or stiff-type landing (324 ms, 286 ms, and 133 ms, respectively) than male soft-type landing technique (334 ms, 223 ms, and 117 ms, respectively).[5] Although the time to the peak ankle dorsiflexion was not significant, the duration to the peak knee and hip flexion were significantly shorter. As the result of shorter landing phase, both stiff landing techniques demonstrated greater vertical GRF (3.16 - 3.56 BW) compared to soft landing techniques (2.25 - 3.21 BW).[5,6]

In addition to the difference in the magnitude of vertical GRF between landing types, the time to the peak force were also notable. It was suggested that maximum stress was loaded on the ACL as short as at 40 ms after the initial contact.[36,119] Thus, the shorter time to the peak landing force might contribute to greater GRF and excessive loading on the ACL.

Ankle Range of Motion and Knee Valgus during Landing Tasks

Researchers found that smaller ankle dorsiflexion could increase GRF due to a lack of landing force attenuation.[5-7] By landing on a 3.6° inclined surface, the ankle ROM during landing tasks was modified. During the two-leg drop jump task, ankle dorsiflexion was limited by 3° increased, knee valgus increased by 1.4°, and the GRF increased by 10.2% (BW) compared to landing on a flat surface.[39] Although the clinical importance of 1.4° of knee valgus increasing was not clear due to the lack of similar studies,[39] previous studies reported that 2.5° knee valgus angle could apply significant ACL strain,[120] and that increasing knee valgus angle by 4° produce 15% greater strain in ACL.[121] The author suggested the greater GRF during landing with limited dorsiflexion could increase the load on ACL. Although the neuromuscular activities were not measured in this study, the quadriceps muscle might need to increase activity to counteract the external flexion moment,[39] and this higher force produced by the quadriceps muscle might also contribute to non-contact ACL injury.[122] However, it should be noted that this study[39] measured lower extremity kinematics and kinetics with landing on an inclined surface, but did not limit actual ankle ROM. This result was also supported by squat with a wedge under the forefoot to reduce dorsiflexion ROM so that the ankle dorsiflexion was restricted.[4]

During a descent phase of squat movement, decreased ankle dorsiflexion was associated with greater MKD.[4] The knee valgus during a landing task was believed to increase the risk of non-contact ACL injury.[45,99] Therefore, the ankle ROM, particularly dorsiflexion, in addition to plantar flexor strength, possibly plays a significant role in reducing GRF and frontal plane knee movement, and subsequently the risk of non-contact ACL injury.

Ankle ROM restriction was related to greater MKD in descending phase of squat that is similar to the descending phase of a two-leg landing task. During a two-leg landing, limited ankle ROM increased knee valgus angle that was found to be associated with MKD in addition to greater GRF. Therefore, MKD and GRF could be increased by smaller ankle ROM, particularly dorsiflexion, that may increase the risk of non-contact ACL injury.

Ankle Range of Motion and Risk of Non-contact Anterior Cruciate Ligament Injury

Ankle joint has hardly been focused in studies that investigated a risk of non-contact ACL injury. Previous studies showed the ankle ROM, the strength of plantar flexors, and landing types might be associated with non-contact ACL injury. During a landing task, the lower extremity must absorb a landing impact. The importance of plantar flexion strength was explained by its role as a shock absorber against a landing impact. The ankle joint could attenuate approximately 80% of landing force[6], and greater ankle displacement demonstrated larger ability to absorb landing force.[3,7] The landing force that was not absorbed at the ankle joint would be transferred to the proximal joints including the knee and the hip joint.[108] The knee joint and the hip joint must have greater angular displacement to dissipate the transferred force. However, if a weaker gastrocnemius was unable to rotate the tibia internally during the knee flexion, the medial knee displacement would increase.[3,4] Therefore, the link between less landing force attenuation ability at the ankle joint and ACL injury was filled by insufficient strength against quick dorsiflexion movement and subsequent greater GRF, and greater medial knee displacement due to weaker gastrocnemius. Future research should also investigate the ankle kinematics and kinetics to assess a risk of non-contact ACL injury in addition to the knee and hip joint kinetics and kinematics.

Greater ankle ROM along with greater peak angular velocity and higher rate of load attenuating function at the ankle joint among females[6,38,116] indicated that smaller ankle ROM in the sagittal plane and/or plantar flexor weakness, especially in the soleus, might increase the risk of non-contact ACL injury. These gender differences could increase stress on the ACL of female and be a potential factor associated with the gender difference in non-contact ACL injury rate.

Sagittal Plane Knee Joint Kinematics and Kinetics

The residual force that was not absorbed by the ankle joint could be transferred to the knee joint. The association of greater GRF and subsequent excessive knee valgus could be explained by ligament-dominant landing technique, which might increase the risk of non-contact ACL injury.[30] When the ankle joint absorbed the landing force enough along with less knee flexion during a landing task, the GRF was similar to a natural landing with more knee flexion and less ankle plantar flexion.[3] When instructed to land with less knee flexion or to land on heels, both GRFs were similarly greater than the knee flexed natural landing and the plantar flexed with less knee flexion landing.[3] Similar to the ankle joint sagittal plane ROM, the knee flexion angle during a landing task was inversely related to GRF.[110] The results indicated that not only greater displacement of the ankle joint but also greater knee displacement could contribute to reduce GRF during a landing. The less GRF is believed to decrease the loading stress on the ACL; therefore, the risk of non-contact ACL injury might be reduced, as well.[82]

Previous studies often investigated knee joint kinematics and kinetics during landing tasks in conjunction with gender differences. The results showed males often attenuate landing impact more effectively than females.[6,38] However, contribution of the sagittal plane knee angles to the magnitude of GRF has been controversial. Females tended to land with less knee flexion and demonstrated greater GRF than males (females = 30 N/J; males = 19 N/J) during single-leg hopping.[116] The GRF was reduced significantly with a greater knee flexion angle at initial contact phase of a single-leg landing task.[123] When the initial contact was at less than 25° of knee flexion, the resultant GRF was 19.3 N/kg. This GRF was significantly larger than the landing with 65° knee flexion at the initial contact that recorded less GRF of 16.4 N/kg.[123]

During a two-leg drop jump task, males and females demonstrated similar initial contact knee flexion angles (male = 14.87°; female = 14.83°) and maximum joint angles (male = 88.91°; female = 87.25°). Females, however, demonstrated greater peak vertical GRF and posterior GRF than males possibly because females activated lower extremity muscles to resist landing force before and during the landing different from males.[38] Another study found different results. When female landed with less initial-contact knee flexion (males = 30.0°; female = 22.8°) but greater maximum flexion (males = 33.4°; females = 53°) than males, no difference were observed in normalized GRF (males = 3.67 BW; females = 3.39 BW) and time to peak GRF (males = 40.00 ms; females = 44.38 ms) between genders.[115] Although both studies assessed landing from a 60 cm height, the first study used vertical drop from a hang bar, and the latter study used drop off from a box.

Studies of two-leg landing tasks did not agree with the results of the single-leg tasks. Even though the knee flexion angles at initial foot contact and flexion at maximum flexion were similar, females showed greater GRF than males. Although, the smaller knee flexion angle at the initial contact might be compensated by greater peak knee flexion angle to attempt to maintain GRF relatively consistent, the insufficient muscular strength could not resist to the landing force. As a result, females demonstrated greater GRF compared to males. Also, different types of landing might induce different foot contact strategies and subsequent knee joint kinematics between tasks. In other words, vertical landing task might induce different landing strategies than forward landing in terms of initial foot contact.

Although the association between the sagittal plane kinematics and GRF was inconclusive, the rate of loading could affect GRF during landing tasks. When looking at GRF loading rate, the greater GRF loading rate increased the GRF during landing tasks. The results elucidated that females had lower capability of landing force absorption compared to males during landing phase in a single-leg hop. During a single-leg hop landing, the standardized GRF loading rate was greater in females (260 N/Js) than males (190 N/Js), which was approximately1.5 times larger in females than males.[116] However, these results were not supported by another study that examined the angular velocity at the knee joint.

A study observed a greater angular velocity in a knee joint did not result in a larger maximum GRF.[115] During a drop landing task a 60 cm box, female exhibited greater ankle and knee displacements in the landing phase, but the peak angular velocities at the ankle and the knee joint were also greater in females (ankle = 1044°/s; knee = 725°/s) than males(ankle = 573°/s; knee = 602°/s). The study, still, did not found a significant difference in peak GRF between genders (females = 3.67 BW; males = 3.39 BW).[115] Although, the loading rate might not be associated with angular velocity and be considered separately, the two studies also used different tasks. The one study used 15-second repeated in-place hopping task[116], and the other used 60-cm two-foot drop jump task.[115] In addition to possible muscular activation strategies, the task difference might produce different results.

Another study of single-leg landing from a 30 cm box also reported that knee flexion angular velocity was less in females, but normalized GRF was significantly greater in females.[6] The initial-contact knee flexion angles were similar in males and females, but the angular displacement at the knee joint was less in females. The time to peak flexion was also shorter in females (223 ms) than males (285 ms possibly due to smaller angular displacement. The females had slower mean angular velocity (37.2°/s) than the males (45.3°/s), the GRF was greater in females (3.56 BW) than males (3.21 BW).[6] These results might imply that although the height of the platform or the types of landing (single-leg or double-leg) could be interacted, males and females used different strategies in landing tasks. In other words, females landed with greater angular velocity in the ankle joint, but smaller angular velocity in the knee joint. In contrast, males landed with smaller angular velocity in the ankle joint, but greater angular velocity in the knee joint. Muscle activities of the lower extremity, still, need to be evaluated to study this relationship to conclude greater GRF combined with smaller peak knee flexion angle might influence on the risk of non-contact ACL injury.

Frontal Plane Knee Joint Kinematics and Kinetics

The risk of non-contact ACL injury is believed to be related to knee valgus angle during landing. Although analyses of video recordings found that ACL-injured subjects landed with greater knee valgus and increased valgus angle progressively, the knee valgus angle was actually fluctuate during a landing task. Still, greater knee valgus angle during a landing task may increase the risk of non-contact ACL injury, particularly immediately after initial foot contact. This greater knee valgus angle was also associated with smaller peak knee flexion angle, which was correlated with greater GRF and may increase the risk of non-contact ACL injury.

Three-dimension kinematics of non-contact ACL injury was reconstructed from the video recordings of non-contact ACL injury moments. The ACL-injured individuals demonstrated rapidly loaded on ACL up to 40 ms after initial contact by increasing knee valgus angle along with internal rotation of the knee joint.[36] Therefore, the frontal plane kinematics could increase non-contact ACL injury risk. Knee valgus during landing tasks should also be considered as a non-contact ACL injury factor.[45] A non-contact ACL injury two-dimension video analysis revealed that the knee valgus angles were similar in ACL injured (5.5°) and non-injured individuals (5.6°) at the initial contact; however, the ACL injured subjects demonstrated progressively increased the knee valgus angle (37.7°) compared to non-injured group (9.0°) during a landing.[37] This knee valgus motion was believed to be the result of ligament dominance landing.[30] It must be noted that the moment of ACL rupture was not determined by the video analysis; therefore, the greater knee valgus angle might be the result of an ACL tear.[37] Still, in these video analyses, capturing accurate joint angles was difficult due to the low frame rate of general broadcasting on television.

During landing tasks, women tended to land with more frontal plane knee movement and less sagittal plane knee movement than men. Females demonstrated greater knee valgus at initial contact of the foot than males during single-leg drop landing tasks.[31,86] During a landing followed by two-second "stick," the knee frontal plane angles of females and males at the initial foot contact were 0.65° valgus and 3.9° varus, respectively. At the peak knee flexion, these angles were 3.1° varus in females and 15.3° varus in males.[86] Although female landed with slight knee valgus, both females and males increased the varus angle as the knee flexion increased. The authors suggested that the sagittal plane angle could be more associated with frontal plane knee angle than with a gender difference; hence, the association had to be examined if each degree of knee flexion is different between genders.[86]

Dynamic change of frontal plane knee angles were reported when the knee frontal plane movement was analyzed after initial contact (up to 500 ms) during a single-leg medial and lateral landing. The result exhibited that the dynamic change of knee frontal plane angles would not have a correlational relationship with the knee flexion angle.[31] The knee flexion angle would continuously increase until the peak knee flexion angle is reached; however, the frontal plane knee angles actually fluctuated during the single-leg landing tasks. In both medial and lateral landings, the knee valgus angle progressively decreased immediately after the initial contact (up to 50 ms). Following the first varus motion, the knee valgus angle quickly increased (up to 100 ms), and this knee valgus angle was recorded as the peak knee valgus. The knee valgus angle was, then, rebounded back to neutral position, and the dynamic change of frontal plane knee movement was plateaued or gradually increased.[31] This result also supported that a ACL tear might occur within 40 ms after the initial contact.[36] Following the initial reduction of the valgus angle, the knee suddenly suffered the peak knee valgus angle regardless the direction of landing.

Even though the frontal plane knee movement showed an oscillating movement, females often demonstrated a greater maximum valgus angle at the knee joint compared to males regardless landing tasks.[32,38] Single-leg forward landing resulted in significantly greater knee valgus in females (7.26°) than males (3.29°) without demonstrating any differences in hip flexion, hip adduction, hip internal rotation, and knee flexion.[32] During a two-leg drop landing from a 60-cm hang bar, females showed greater frontal plane knee movement (26.5°) compared to males (7.1°). The authors found that the greater difference in the knee valgus angles between genders occurred at the first 30 to 50% and the last 10% of the landing phase.[38] Therefore, not only immediately after the initial foot contact, the frontal plane knee angle at the peak knee flexion should also be analyzed.

Frontal plane knee motion was examined with two groups divided by the value of the total knee and hip flexion, and the landing with greater hip and knee flexion could result in the lower knee valgus during a two-leg drop landing task.[40] The high-flexion group recorded the total flexion angle above the mean of all subjects (peak knee flexion = 100.6°; peak hip flexion = 89.9°); the low-flexion groups showed the total sagittal plane angle below the mean (peak knee flexion = 86.5°; peak hip flexion = 67.4°). The low-flexion angle subjects demonstrated greater maximum knee valgus angle (6.3°) compared to the high-flexion angle subjects (3.9°).[40] The results implied that a landing with a low sagittal plane flexion might result in greater valgus loading. This high valgus loading landing technique was also found among non-contact ACL injured adolescent females.[30,37]

A prospective controlled cohort study revealed that during a drop vertical jump task, ACL injured subjects demonstrated a landing technique with greater knee valgus and smaller knee flexion angles that was believed to increase non-contact ACL injury risk.[99] Three joint angles that showed significant difference between groups were knee valgus angle at initial contact, knee valgus angle at maximum knee flexion, and maximum knee flexion angle. The ACL injured group showed greater knee valgus (5.0°) at the initial foot contact compared to the uninjured group (3.4°). The knee valgus angle increased in the ACL injured group (9.0°), but the knee valgus reduced in the uninjured group (1.4°) during a descending phase of the drop vertical jump task. Moreover, the maximum knee flexion was also smaller in the ACL injured group (71.9°) than the uninjured group (82.4°).[99] Therefore, the frontal plane knee joint angle might be related to the sagittal plane knee joint angle during non-contact ACL injury.

There was a consensus that greater knee valgus during a landing task could increase the risk of non-contact ACL injury[35-37] possibly due to higher ACL loading that might attribute to the landing technique with smaller knee flexion and greater knee valgus.[2,30,99] Moreover, the study that compared the ACL injured and uninjured groups found an association between knee valgus angle and peak GRF. The ACL injured group with greater knee valgus motion and smaller knee flexion demonstrated greater vertical GRF (1266 N) compared to the uninjured group (1058 N) during the drop vertical jump task. A significant correlation was observed between the knee valgus angle and the maximum vertical GRF only in the ACL-injured group (R = .67).[99]

The knee valgus angle has been believed to increase the ACL loading during a landing task.[37,45,99] The greater knee valgus angle was possibly associated with the smaller knee flexion angle[40,99] and smaller ankle dorsiflexion[4,39] in the descending phase of landing. The force transmitted to the proximal joints was a residual force that was not absorbed in the distal joint. Therefore, the ankle joint kinematics and kinetics could play significant roles in ACL loading force in the knee joint. Role of the ankle joint on the risk of non-contact ACL injury has not been sufficiently investigated to date. Future studies need to investigate the association of the ankle joint energy absorption and the risk of non-contact ACL injury.

Greater knee valgus angle are likely to increase the risk of non-contact ACL injury. This association was supported by analyzing video recordings of ACL injury. Studies[35,36,80] found that the knee valgus angle immediately after initial foot contact could lead to greater peak knee valgus angle. Therefore, a landing with greater frontal plane angles may increase the risk of non-contact ACL injury. However, the association between the knee joint kinematics and the ankle joint kinematics has hardly studied thus far.

Landing Tasks and Landing Biomechanics

In addition to types of landing (soft and stiff landings), tasks of landing also influenced the kinematics and kinetics. Two commonly used landing tasks to assess non-contact ACL injury risk were drop-landing,[3,5,86,119,124] and forward-jump landing.[113,124] A study comparing the two landing tasks concluded that the two tasks were different. Forward-jump landing involved concentric and eccentric muscle contraction during the task; on the other hand, drop-landing involved only eccentric phase of muscle contraction.[82] Although the ankle joint did not show asymmetry between two legs during the two landing tasks, the forward-jump landing task exhibited greater asymmetry at the hip adduction angle and the knee valgus angle in frontal plane kinematics than drop-landing task. This result indicated that the asymmetry in knee valgus might be associated with the greater risk of non-contact ACL injury during forward-jump task which includes landing and deceleration.[2,82]

Although two different types of landing tasks have been examined to assess the risk of non-contact ACL injury, only one study[82] investigated these two landing tasks. It was concluded that female subjects demonstrated asymmetrically greater valgus angle in a dominant leg than males during a forward-jump landing. The kinematics of ankle and knee joints have not been studied in two different types of landing tasks yet. Therefore, it is also important to compare the two types of landing tasks in terms of available ankle range of motion (ROM).

External Ankle Supports

External ankle supports (EAS), such as ankle taping and ankle bracing, have been frequently used to prevent ankle sprains in athletic settings. Although ankle taping was often used in athletic training, re-usable ankle lace-up brace and semi-rigid brace have become more common because of high cost of daily taping and irritation to the skin by tape adherent or adherent spray.[125] The goal of the EAS is restricting ankle frontal plane movement, such as inversion and eversion, at the subtalar joint. However, many of EAS inevitably limit ankle sagittal plane movements, plantar flexion and dorsiflexion, which are essential movements during athletic movements. Therefore, EAS may not only affect athletic performance but also alter landing mechanics due to limited plantar flexion and dorsiflexion. The influence of limiting ankle sagittal plane movements during functional performance have been debatable; still, changing landing kinematics and kinetics might contribute to increasing a risk of non-contact ACL injury.

External Ankle Supports and Ankle Function

Ankle taping and lace-up ankle braces successfully decrease ankle inversion and eversion ROM. The EAS effectively limit inversion ROM (12 – 47%)[125]; hence, EAS decreased total ROM in the frontal plane at the ankle joint.[8,41,125,126] Although these studies demonstrated EAS effectively restricts the frontal plane ROM, the effect of ankle injury reduction was still questionable.[125,127]

Ankle inversion sprain frequently occurs when the ankle is in slightly plantar flexed position and in inversion.[44] Consequently, limiting plantar flexion should be one of the purposes of the use of EAS. Ankle taping techniques and lace-up ankle braces efficiently reduced passive (59%)[128] and active (26%)[129] ankle plantar flexion ROM,[41,128,129] but the dorsiflexion was also limited (50.0%)[128] as the result of the EAS application.[8,41] Compared to non-taped condition, the ankle taping decreased the plantar flexion angle (non-taped condition = 65°; taped condition = 26.5°) and reduced the dorsiflexion angle (non-taped condition = 20°; taped condition = 10°).[128] As a result, the total sagittal plane ankle ROM was reduced.[8,125,126] This limited plantar flexion and dorsiflexion might contribute to alter functional performance during running and jumping tasks, and alter proximal joint kinematics during various landing tasks.

External Ankle Supports Do Not Influence Functional Performance

Although previous studies revealed prophylactic EAS effectively restricted ankle ROM in sagittal and frontal plane, constricted sagittal plane ROM also interfered plantar flexion and dorsiflexion that were essential movement during functional tasks, such as jumping and landing tasks. Studies concluded that EAS hardly affected or had no negative effect on performance during functional tasks. EAS reduced sprint speed by only one percent, and the reduction was not significant.[8] Agility speed reduced up to 0.5% by semi-rigid ankle brace application, and ankle taping or lace-up brace did not affect agility performance.[8] Single-leg static balance was not different between no EAS and EAS groups.[130,131] Application of EAS did not change athletic performance in vertical jump height or broad jump distance even though the sagittal plane ankle ROM was limited.[8,41,129,131,132] Many researchers had hypothesized that application of EAS, such as ankle taping, lace-up brace, or semi-rigid brace, would deteriorate functional task performance, especially vertical jump height. The results did not support their hypotheses possibly due to measuring only static open kinetic ROM.[132] In other words, limited ankle ROM measured in open kinetic chain might not have influence in dynamic ROM during functional tasks. In addition, long-term effects of EAS were not observed; hence, wearing EAS for months or years might influence on muscular strength or joint proprioception and on functional task performance.

External Ankle Supports Alter Landing Kinematics

Although EAS did not negatively affect functional performance, such as sprinting speed, agility speed, balance, or vertical jump height, they altered kinematics of the lower extremity during landing, side cutting, and squatting tasks. Kinematic changes due to altered ankle ROM were observed not only in the ankle joint but also in the proximal joints in the lower extremity. The reason for this change was possibly because the movement of the distal segment influenced the proximal joint through the closed-kinetic chain system.

Application of EAS decreased the plantar flexion angle at ground contact (no EAS = 24.5° – 37.8°; EAS = 16.9° - 34.4°).[133,134] The maximum dorsiflexion angle during a landing phase was also reduced by EAS (no EAS = 23.2° - 21.5°; EAS = 21.0° - 22.1°).[134,135] As a result of less plantar flexion before landing and limited ankle dorsiflexion during landing phase, the total ankle joint displacement was also decreased.[134,135] This reduced range of ankle motion could lead to decreased time to peak GRF during drop landing tasks during a 60-cm box drop landing task.[136] Still, another study revealed that the time from the foot contact with the ground to the maximum dorsiflexion did not change during a 30-cm height two-foot drop-landing and immediate vertical jump task.[134] These inconsistent results might possibly be due to the use of different heights of landing (60 cm vs. 30 cm) and types of EAS (semi-rigid ankle brace vs. lace-up ankle brace).

Ankle motions in the frontal plane including inversion and eversion did not change by ankle tape application during a single-leg drop landing task,[133] but the frontal plane movements were effectively reduced in dynamic side-cutting tasks.[137] Thus, during a single-leg or two-leg landing task that primarily involved sagittal plane motion, EAS likely to affected only sagittal plane kinematics.

Limited ankle sagittal plane motion by EAS might also change the rotation of the tibia and subsequent joint kinematics during landing or functional tasks. Orthotics application to the females with pes planus did not change tibial rotation angle,[138] but wearing EAS decreased the tibial internal rotation during running, cutting, and crossover cutting tasks,.[137] This indicated that simply supporting the medial longitudinal arch did not change landing kinematics, but stabilizing the ankle joint by EAS could affect the lower extremity kinematics during functional tasks.

It was understandable for EAS to alter the kinematics at the knee joint during landing task. EAS changed the knee joint angles during a forward-jump landing task but did not affect during a simple vertical drop landing task. Application of lace-up brace increased the knee flexion angle at the foot contact but did not change the peak knee flexion angle; as the result, the knee sagittal plane displacement decreased. The knee joint sagittal motion appeared to be compensating the ankle-stabilizer-induced reduction of the ankle sagittal motion by increasing knee flexion at the foot contact so that the GRF were relatively kept constant.[134] The time to peak knee flexion did not change, and this result was comparable to the time to peak dorsiflexion angle.[134] Another study used a hanging vertical drop landing demonstrated no difference in sagittal plane knee displacement among control, ankle taping, and lace-up brace.[139] This disagreement could attribute to the different landing methods; the former used a forward-jump landing, and the latter used a simple vertical drop landing.

Regarding the sagittal plane knee angle, the knee flexion angle and the anterior tibial shear force were inversely associated; therefore, less knee flexion during a landing task increases the anterior tibial shear force. Thus, application of EAS and reduced sagittal plane ankle displacement could reduce the tibial anterior shear force during the forward-jump landing task.[134] This reduction of sagittal plane ankle displacement could result in greater peak landing force because the ankle dorsiflexion ROM and the plantar flexors played an important role in landing force attenuation.[5,113]

The sagittal plane ankle ROM measurement was associated with the knee and the hip displacement. The dorsiflexion angle measured in knee extension was inversely correlated with the knee sagittal plane displacement and the hip sagittal plane displacement during a landing task. Therefore, in addition to the ankle ROM, the gastrocnemius muscle activation or strength could be related to the GRF.[113] Another correlational study showed that the less dorsiflexion angle was associated with the greater dynamic valgus.[140] Furthermore, restriction of the ankle dorsiflexion by placing a wedge under the forefoot during the controlled squatting exercise simulated the restricted plantar flexion flexibility.[141] It should be noted that those correlational studies did not use EAS to limit ankle ROM, and the use of EAS and proximal joint kinematics during landing task have not been studied to date.

External Ankle Supports Alter Landing Kinetics

EAS affected the proximal joint kinematics during landing tasks. The different lower extremity kinematics could also influence the kinetics during the same tasks. The kinetics measured were vertical GRF, time and rate to the peak GRF, muscular activities (the vastus medialis oblique, vastus lateralis, gastrocnemius, soleus, peroneal muscles, and tibialis anterior), and time to stabilize after a landing task. The results of GRF and muscular activities were not consistent probably due to different tasks performed.[41,134,136,137,139,142-145]

The maximum GRF did not change during forward-jump landing tasks[134,142] and dynamic cutting tasks[137] following the application of EAS. When the landing was divided into two phases, toe contact (first peak GRF) and heel contact (second peak GRF), the magnitude of the GRF depended on the task performed. The GRF significantly decreased both at toe contact and at heel contact during two-leg drop landing after EAS application.[136] In contrast, during hanging drop landing, EAS application increased GRF at the toe contact phase but did not change at the heel contact phase. As a result, the time to peak GRF decreased at the toe contact; hence, the rate to the peak GRF also increased.[139] During basketball-type side cutting drill, wearing EAS did not change GRF at the toe contact phase but decreased maximum GRF. Moreover, EAS application decreased the time between the first peak and the second peak GRF; therefore, the rate of GRF increased.[143]

EAS application modified muscular activities in the lower leg muscles. During a single-leg forward-jump landing task, the gastrocnemius and the peroneus longus activities decreased, and the tibialis anterior activity did not change in EAS group compared to control group.[142]

Although it was not measured, reduced calf muscle activities could indicate that the mechanical ankle stability might decreased during the landing task. Other studies[41,144,145] demonstrated that the peroneus longus and brevis activities were not changed by EAS application. However, these studies measured muscular activities during sudden inversion[41,144] or transient oscillations[145] instead of sagittal plane landing tasks. Thus, the disparity of peroneus longus activity could attribute to the tasks performed, and the EAS application might reduce the activity of the peroneus longus during the sagittal plane landing task.

Restricted dorsiflexion could alter the soleus muscular activation level because the soleus muscle eccentrically resist ankle dorsiflexion to control sagittal plane ankle motion.[141] When the knee is flexed the gastrocnemius loses its proper tension to generate force. In contrast, the soleus muscle length-tension relationship is not influenced by the knee flexion. Consequently, the soleus activity level could increase during the deceleration phase of squat-type movement. In drop-landing tasks, the soleus activity possibly becomes more important than in forward-landing tasks. This is probably due to the necessary displacement of the ankle joint between the tasks.

Non-contact Anterior Cruciate Ligament Injury Risk Assessments

Three screening assessment techniques, including tuck jump, drop jump, and landing error scoring system (LESS), have been commonly used by utilizing two-dimensional commercial digital camcorders to evaluate potential non-contact ACL injury factors of sagittal and frontal plane kinematics. Because the gold standard three-dimensional movement evaluation is costly and time-consuming, field-based movement assessment tools using commercially available digital camcorders are used by clinicians and coaches. By using two-dimensional video images recorded at 30 or 60 Hz, all of those assessment tools successfully evaluated lower extremity kinematics difference between males and females during landing phase of various landing tasks.

Tuck Jump and Drop-Jump Assessments

Tuck jump assessment tools were developed to help detect landing errors by observing knee and hip flexion angles, lower extremity valgus angle, asymmetry of extremities, foot placement, foot contact timing, and landing noise.[2,146] In the tuck jump assessment, participants performed 10-second repeated tuck jumps, and the clinician visually rated the landing technique based on the grading criteria. Two standard camcorders were placed in front and on the side of the landing platform to obtain kinematics of frontal plane and sagittal plane, respectively. The two main focuses of the tuck jump assessment is dynamic knee collapse and side-to-side asymmetry during the task.[146] It should be noted that the lower extremity external and internal rotations of the femur and tibia could affect the frontal plane apparent motion at the knee joint, and the frontal plane movement analyzed by two-dimensional video recording system was not representative of true knee valgus.

Drop-landing vertical jump has been used to evaluate the lower extremity frontal plane dynamic knee valgus collapse during landing phase and accelerating phase.[47,140] The drop-landing vertical jump consisted of jumping off a box of 31cm[47] or 46cm[140] in height, natural landing, and immediate maximum vertical jumping.[47] The knee valgus collapse was evaluated at the initial toe contact to the ground, at the maximum knee flexion, and the initial forward movement of the body for a vertical jump. In drop-landing studies, the dynamic knee collapse was not measured in angles; instead, the distance between the knees were normalized to the participant's height[140] or to the distance between the left and left anterior superior iliac spines of the hip[47] in addition to actual distance between the hips. Hence, the dynamic knee valgus collapse was expressed as knee separation ratio relative to the hip distance.[47]

The knee separation does not measure the true angle of knee valgus in the frontal plane, but simply means the distance between the knee joints. The knee apparent movement observed by a single-plane video analysis could be the results of combined movements in the transverse and frontal planes. Decreased hip external rotation ROM was associated with the frontal plane knee valgus.[140] Hence, the knee separation measurement implied not only the internal and external rotation of the femur and tibia, but also internal rotation of the hip joint. The test-retest reliability of the hip separation distance was found to be substantially high (ICC $\geq$ 0.94), and the within-test reliability for the hip, knee, and ankle separation distance were substantial (ICC $\geq$ 0.90).[47]

Forward-Jump and Landing Error Scoring System

Landing Error Scoring System (LESS) is a newly developed jump-landing movement evaluation tool to identify landing errors by reviewing two-dimensional recorded images with video analysis software programs.[147] Based on jump-landing characteristics, the LESS simply used a dichotomous scoring system to several items. The LESS measured knee flexion angle, knee valgus angle, trunk flexion angle, ankle plantar-flexion angle, foot/ toe position, stance width, foot contact symmetry, sagittal ROM in all lower extremity joints, and overall quality of landing.[148] The LESS was originally proposed to assess potential landing errors associated with non-contact ACL injury; therefore, the scoring system was more comprehensive landing assessment tool than previous clinical assessment tools.[46]

Similar to the drop-landing vertical jump task, the LESS also involved landing and vertical jump tasks. However, a key difference in LESS was that the LESS required horizontal motion in addition to landing movement which was often involved in non-contact ACL injury.[46] The suggested task consisted of a two-leg forward, but not vertical, jump from a 30cm-height box, two-leg landing on to the platform at a distance of 50% of subject's body height, and immediate maximum vertical jump.[46,149] The investigators instructed the subjects to complete the task as naturally as possible.[150] Recording of the initial landing of the box was reviewed in the sagittal and the frontal plane by a rater using the LESS, and the total error score would be calculated.[151] Each faulty landing characteristics was scored as an error, and the higher LESS score indicate poorer landing technique.[148,150]

The LESS is a valid and reliable tool to characterize erroneous landing technique during a forward-jump landing task to identify with high non-contact ACL injury. The LESS had excellent intra-session ($ICC_{2,1} = 0.90$), inter-rater ($ICC_{2,1} = 0.84$) and intra-rater ($ICC_{2,k} = 0.90$) reliability in a drop-landing task.[151,152] More importantly, the inter-rater ($ICC_{2,k} = 0.84$) and intra-rater ($ICC_{2,1} = 0.91$) reliability for the LESS was also excellent in the forward-jump landing task. The scoring system successfully differentiated gender differences and high/low erroneous groups[151]; however, the validity of the LESS appeared to be item dependent when it was compared to three-dimensional motion analysis.[152]

When the scores of experienced-raters were compared to the three-dimensional motion analyzing tool, excellent agreement (> 84%) was shown for ankle flexion angle at initial contact, total knee flexion ROM, trunk flexion at maximum knee flexion, and foot position (tibial rotation) at initial contact. Moderate agreement (68 – 74%) was demonstrated for trunk flexion at initial contact, stance width relative to the shoulder width, knee valgus at initial contact, and total knee valgus ROM.[152] Significant correlation with three-dimensional motion analysis (κ = 0.46 – 1.0) was also observed for total knee flexion ROM, symmetric initial foot contact, foot position at initial contact (toe-out), narrow stance width (< shoulder width), knee valgus at initial contact, lateral trunk flexion at initial contact, total knee valgus ROM, and total sagittal plane ROM in the drop-landing task.[152]

The higher LESS score with erroneous landing technique was associated with less flexion angles at the knee and hip, greater knee valgus, greater hip adduction, greater anterior tibial shear force, and greater GRF in the forward-jump landing task.[46] The important risk factors for non-contact ACL injury were knee flexion angle, knee valgus angle, and hip adduction angle; therefore, the LESS is an excellent assessment tool to identify landing errors that might contribute to non-contact ACL injury.

The gold standard three-dimension motion analyses usually use a high-speed filming technology between 150 and 500 Hz.[113,133,152] A drawback of two-dimension video analyses of tuck jump, drop jump, and LESS to evaluate landing technique were incapability of commercially available camcorder to capture kinematic data.[152] Those camcorders usually capture images 30 or 60 Hz[45,152,153]; therefore, high-speed human movements, especially in the sagittal plane movement, were probably not clearly pictured when raters reviewed and evaluated the landing techniques. Even using a three-dimension motion analysis to analyze human movement, filming 60 Hz was not appeared to be sufficient to capture detail of joint angles during a drop-landing task.[154]

Summary

The risk of sports-related injury is inevitable in physical activity, particularly in sports. Interestingly, non-contact ACL injury is substantially more prevalence in females than males. Potential factors and theories have been proposed; however, no definitive cause has been identified, and it is likely that multiple factors could contribute to this phenomenon. One of the factors identified by video analyses was excessive knee valgus angle and greater GRF during a landing, and this was supported by other experimental researches. Adolescents change their strength and postural characteristics during their puberty period, and postpubertal females, in general, demonstrate greater knee valgus angles during landing tasks. Because landing force that is not attenuated in the ankle joint has to be transferred to the proximal joints, ankle joint ROM and strength, particularly plantar flexors, are essential during a landing. However, the relationship between ankle ROM and kinematics in the proximal joint and ground reaction force (GRF) is still unknown. The ankle ROM in frontal plane is successfully restricted by external ankle supports (EAS) in athletic settings, but EAS also limited the sagittal plane ankle ROM. Hence, the influence of EAS regarding knee kinematics and kinetics during landing tasks needs to be identified.

METHODS

Purpose of study

Females are more at risk of a non-contact ACL rupture than males. One cause is female's landing mechanics differs from their male counterparts. Less flexion of the lower extremity during a deceleration phase of activities and subsequent greater GRF and knee valgus might increase the risk of the ACL injury. Drop-jump landing (DJL) and forward-jump landing (FJL) tasks are inevitable in many sports activities. Therefore, the first purpose of this study was to investigate whether the limited ankle ROM by application of external ankle supports (EAS) changes landing kinematics and kinetics. Therefore, the second purpose of study was to investigate whether the two types of landing tasks differ in kinematics and kinetics.

Experimental Design

To assess the effect of EAS, landing tasks, and isokinetic strength of the soleus, a 3 x 2 factorial crossover-repeated measure design was used. The independent variables were EAS conditions and landing tasks. Three EAS conditions tested were no ankle braces (NB), ankle braces with low-tension (LTB), and ankle braces with high-tension (HTB). The NB was a comparison group. Two types of landing technique performed were drop-jump landing and forward-jump landing. The order of EAS conditions and landing tasks were counterbalanced by using a Latin square and crossover, respectively. The dependent variables were static ankle plantar flexion and dorsiflexion with knee flexion, static ankle dorsiflexion with knee extension, static ankle dorsiflexion with weight bearing, ankle displacement during a landing, peak ankle dorsiflexion during a landing, peak medial knee displacement (MKD) during a landing, peak vertical GRF during a landing, concentric peak torques of plantar flexors at 60°/s, 120°/s, and at

180°/s, and eccentric peak torque of plantar flexors at 30°/s. In addition to age, height, and weight were measured, the day of last menstrual cycle was also asked.

Sample Size and Sampling Method

Based on the data obtained in pilot studies, the minimal number of subjects in each condition required with a power of 0.8 and a level of significance α of 0.05 was calculated[155] to be 14.9. Convenience sampling method was used, and physically active healthy adults were recruited from HNES 210 (First Aid and CPR), HNES368 (Biomechanics of Exercise), and HNES 365 (Kinesiology) class. Potential participants were approached by in-class recruitment and e-mail. Criteria to participate in the study included no neuromuscular dysfunctions, no history of prior surgeries within six months, or signs and symptoms of inflammation including pain and/or joint effusion in the lower extremities. A physically active healthy college-age female was defined as a person who was between 18 and 25 years old, who performed physical activity at least two times per week and minimum of 30 minutes in each session, and who had no lower extremity at the time of participation or no history of surgery within six months prior to the participation.

Potential Errors and Bias

There were possible errors in this research project; fatigue effect, order effect, learning effect, selection bias, and instrumentation effect. To minimize fatigue effect, a minimum two minute rest period was provided between each trial. To evenly distribute the order effect, the order of treatments (NB, LTB, and HTB) was counterbalanced by use of Latin square. In addition, the order of landing tasks (DJL and FJL) was alternated. To minimize learning effect, sufficient familiarization period (practice time and repetition) was given prior to testing trials; however, to examine a natural landing, only minimum instruction was provided. To minimize selection bias in terms of pubertal maturation, participants were at least 18 years old at which no further pubertal growth was expected.[156]

Ankle range of motion (ROM) was manually measured by a goniometer. To minimize error regarding goniometric measurement, measurements were taken by the co-investigator who was a health professional with seven years of year experience as an athletic trainer. Although kinematics is commonly analyzed by a three-dimension motion analysis package, sagittal plane ankle kinematics and frontal plane knee kinematics were measured by a commercially available two-dimension digital video cameras. Because joint rotations are not able to be analyzed by a two-dimension video analysis, medial knee displacement (MKD) relative to the distance of the anterior superior iliac spines was assessed, instead. The landing motion was recorded with high-speed video filming mode.

Equipment Set Up

Two digital cameras (EX-FH20; Casio, Inc., Tokyo, Japan) were used to record landing kinematics. One digital camera was mounted on a tripod 73 cm from the floor. The tripod was placed 175 cm in front of a box from which participants took off. This digital camera recorded medial knee displacement (MKD) in frontal plane. Another tripod with a digital camera mounted on was positioned at 175 cm on the right side of the box. This second digital camera recorded ankle displacement in sagittal plane. The box was 33 cm (13.0 inches) in height and 53 cm (20.9 inches) in width. The frequency of filming was at 210Hz which was more than three times sharper or distinct than commercially available digital video recorders. Although the frequency of the recording was more than three times greater, the brightness of the recording was sacrificed; therefore, additional 8-inch studio floodlight (PL8; Smith-Victor Corp., Bartlett, IL) was placed behind each digital camera to additionally illuminate the laboratory lighting.

Participants were instructed to wear athletic shorts and low-cut sports shoes. For frontal plane kinematic recordings, markers were placed at both left and right side of the body; at 2 cm anterior to the acromioclavicular joints, at the anterior superior iliac spine, at the center of patella, and at the middle of the distal lower extremity 5cm superior to the ankle joint. For sagittal plane kinematic recordings, markers were placed only on the right side of the body; at 2 cm lateral to the acromioclavicular joint, at the greater trochanter, at the center of the knee joint line, and at the lateral malleolus.

The force platform (AccuPower; AMTI, Watertown, MA) was embedded to the testing surface. The position of the 33-cm box was adjusted depending on the task performed. For DJL task, the box was placed 5cm from the edge of the platform, which was 35 cm (13.8 inches) from the landing target on the platform. For FJL task, the box was placed at 50% of the participant's height from the center of the force platform. Therefore, participants targeted to land approximately in the center of the platform in both tasks.

The Biodex Multi-Joint system 4 PRO (Biodex Medical System, Shirley, NY) was used to assess concentric and eccentric isokinetic torque of the plantar flexors, especially soleus muscle. The position of a participant was seated on the dynamometer seat, and the position of the ankle was adjusted according to the participant's body length in order for the axis of ankle movement in sagittal plane was aligned with the axis of rotation of the dynamometer.

Experimental Procedure

The procedure consisted of pre-trial evaluation, warm-up exercises, external ankle supports (EAS) application, ankle ROM measurements, and jump-landing trials. The EAS was applied prior to ankle ROM measurements. The ankle ROM was measured in three different conditions; no brace (NB), low-tension brace (LTB), and high-tension brace (HTB). Following ankle ROM measurements, jump-landing tasks were performed. A minimum of 2 minute rest period was provided between jump-landing trials. The orders of EAS applications and jump-landing tasks were counterbalanced. Two measurements were taken in each condition and task and were averaged for analyses. Following the landing trials, the concentric and eccentric isokinetic strength of the plantar flexors was measured at 60°/s concentric contraction, at 120°/s concentric contraction, at 180°/s concentric contraction, and at 30°/s eccentric contraction. A minimum of 2 minute rest was provided between isokinetic measurements. The entire study took approximately 75 minutes to complete.

Pre-trial evaluation

Participants were reported to the Biomechanics Laboratory in the Bentson Bunker Fieldhouse at North Dakota State University and signed the informed consent. Participants were instructed to wear athletic shorts, T-shirts, and low-cut sports shoes for the testing. A history and current signs and symptoms of lower extremity injury and the amount of weekly physical activity were assessed for eligibility. They were also asked about most recent menstrual cycle. Height and body weight were measured by a conventional stadiometer (seca213; seca, Hamburg, Germany) and weight scale (751KLS; Health o meter, McCook, IL), and weight was used to standardize the isokinetic strength.

Warm-up exercise

Participants were allowed to warm up to prepare the body’s neuromuscular system for the demands of landing tasks. Dynamic warm-up exercise consisted of two minute comfortable jogging on a treadmill, 30 second toe walking, 30 second straight leg kicks, 15 second leg swings, 20 second high knee walk, 20 second butt kicks, 20 second lunge walk, and 20 second skipping.

If a subject did not feel he/she was warmed-up enough, additional warm-up exercise was performed by subject's choice. However, no static stretching was allowed.

External ankle supports application

Prophylactic lace-up ankle braces (ASO® Ankle Stabilizer; Medical Specialties, Inc., Charlotte, NC) were fit on both the subject's ankles based on subject's shoe size (small, medium, and large) as instructed by the manufacturer. During the brace application, participant was sitting in a chair and the lower leg was held in perpendicular to the floor so that the ankle joint maintained in 90°. Each lace was tied with a relatively constant tension at 2.5 – 3.0 kg (5.5 lb – 6.6 lb; lower-tension brace, LTB) or at 7 – 7.5 kg (15.4 lb – 16.5 lb; higher-tension brace, HTB) measured by a hand-held digital scale (Rapala VMC, Vaaksy, Finland). Each lace was held between the target tensions for approximately one second at 90° angle from the surface contour of the ankle braces.

Ankle range of motion measurement

After warm-up exercise, ankle ROM was measured in non-weight bearing and weight bearing positions. The non-weight bearing active ROM of the left and right ankle was measured by a 1-degree-increment transparent plastic goniometer (Patterson Medical/ Sammons Preston; Bolingbrook, IL) in two positions. First, a subject was sitting on a table, and the ankle joint was off the table with full knee extension. Second, the subject was sitting on the edge of a table with 90° hip and knee flexion. The reference landmarks were suggested previously.[157] The fulcrum of the goniometer was placed over the center of the lateral malleolus. The stationary arm was aligned with the lateral midline of the fibula by using the fibular head as reference. The moving arm was aligned parallel to the lateral aspect of the fifth metatarsal at maximum dorsiflexion and plantar flexion. The subject was asked to move the ankle toward the body for dorsiflexion measurement and to move the ankle down or away from the body for plantar flexion measurement. The subject maintained the ankle angle for three seconds so that the co-investigator could read the angle measurement. Two measurements were taken by the co-investigator and were averaged for analyses.

Additionally, weight-bearing lunge dorsiflexion was measured by determining the angle of the lower leg relative to the vertical line.[158] First, a single strip of tape was aligned on a wall perpendicularly and on the floor 90° to the wall. Second, two landmarks on the skin were marked by a non-permanent felt pen. One was at 15 cm (5.9 inches) inferior to the tibial tuberosity on the anterior boarder of the tibia. The other was a line perpendicularly bisecting the posterior calcaneus. Third, a participant positioned the foot with the big toe and the calcaneus on the tape. Fourth, the participant lunged forward until the ipsilateral knee was in contact with a vertical tape on the wall. The distance between the foot and the wall was adjusted to find the point where the knee touched the tape on the wall without lifting the heel off the floor. The heel was held by the co-investigator in order to maintain heel contact. Lastly, when the participant reached the maximum dorsiflexion angle, the co-investigator placed the fluid-filled inclinometer (Baseline Bubble; Fabrication Enterprises, Inc., White Plains, NY) on the mark placed on the tibia and recorded the angle. During the lunge dorsiflexion, pronation or supination of the foot was allowed. Two measurements were taken by the co-investigator and were averaged for analyses.

Validity and reliability of joint range of motion measurements

Although several different methods have been developed to measure joint range of motion, a universal goniometer has appeared to be reliable and most widely used in clinical settings because it is simple to use, noninvasive, and inexpensive.[159] The use of universal goniometric measurement is still believed to be reliable and valid compared to other newly developed methods. Reliability of recently developed digital protractor, electronic goniometer, fluid goniometer, and other devices have been compared to a universal goniometer and three-dimension motion analyses.

The goniometric measurement of ankle joint was difficult to standardize due to its complex motion at the talocrural joint, the subtalar joint, and the inter-tarsal joints.[159,160] The standard error of the mean (SEM) for intra-tester in the fluid goniometer, the electronic goniometer, and the universal goniometer were 0.92 – 1.52°, 0.85 – 1.2°, and 0.85 - 0.99°, respectively.[159] Another study[160] found when a single tester took two measurement, the error of universal goniometric measurements on the ankle dorsiflexion was less (SEM = 2.3°) than when the multiple testers took two measurements with the same subject (SEM = 3.1°). Moreover, the SEM slightly increased when the same tester took multiple measurements at different occasions (SEM = 2.4°).[160] Hence, universal goniometric measurements of ankle dorsiflexion should be taken by a single examiner using multiple measurements at one location. Nonetheless, a universal goniometer showed slightly greater intra-tester reliability than the electronic goniometer for assessing active ROM. The active ankle dorsiflexion and plantar flexion intra-class ICC for the universal goniometer and the electronic goniometer were 0.80 – 0.95 and 0.72 – 0.91, respectively.[161] Although most of goniometric measurements revealed high intra-tester and inter-tester reliability, the inter-device reliability decreased when different goniometric devices were used, and the results were compared.[160,162] Therefore, the results of goniometric measurement using different devices should not be interpreted interchangeably.

Possible reasons of intra-tester errors were the validity of an instrument, the tester's reading of the goniometer, and the tester's judgment of the 'end feel' at the limit of range of motion in passive range of motion.[160] To reduce those potential errors, application of a standardized method has been discussed including using a consistent anatomical landmarks, having a subject set in a consistent position, placing a goniometer at a consistent position, taking multiple measurements and averaging them, and using a same torque applied to a joint.[159,160,163] It was a remarkable that a 5-Nm change in torque at ankle joint lead to a 5° to 10° difference[163] and that the variability of measurements less than 7° were associated with measurement error in ankle dorsiflexion measurement.[159] Visual estimation might be the simplest method to measure a joint range of motion, but the reliability of the visual estimation comparing to universal goniometric measurements for the active ankle range of motion were wide ranging ($ICC_{2,1}$ = 0.32 - 0.94).[164] This wide range of reliability suggested that visual estimation not be used to measure ankle joint active range of motion repeatedly.

Although many methods have been used to measure lower extremity joint angle including digital protractor, electric goniometer, fluid goniometer, and universal goniometer, ankle dorsiflexion was usually measured with non-weight bearing position. The ankle joint must be in dorsiflexed position during landing tasks; thus, an ankle dorsiflexion also needs to be taken in weight-bearing position. The weight-bearing ankle dorsiflexion could be measured during lunge movement.[158] For this test, a patient's studying foot faced perpendicularly against a wall, and the patient lunged the knee toward the wall. With the anterior aspect of the knee contacting on the wall, the foot was moved smoothly away from the wall until the maximum ankle dorsiflexion was reached without lifting the calcaneus away from the ground. The angle of tibial shaft relative to the vertical line was taken by an inclinometer at 15cm inferior to the tibial tuberosity.[158] Because the torque applied to the ankle joint at the weight-bearing position was much greater than that applied by an examiner's hand in non-weight bearing methods, the advantage of this method was that the measurement could more represent the ankle dorsiflexion during functional tasks, such as running, jumping, and landing.[158]

Both intra-tester reliability and inter-tester reliability were excellent, and the skill level and experience of the testers did not affect the reliability of dorsiflexion lunge measurements. Intra-tester reliability of a novice tester with minimal clinical experience was excellent ($ICC_{3,3} = .98$), and that of an experienced tester (nine years of clinical experience) was excellent ($ICC_{3,3} = .98$). Inter-rater reliability was also excellent ($ICC_{2,3} = .97$).[158] However, its validity was not investigated, and the researcher noted that this method did not measure a specific joint range of motion. Instead, this measurement was a result of a combined movement at talocrural, subtalar, and midtarsal joints.[158] It was suggested that the result of dorsiflexion lunge measurement not be compared to other methods because this method was measured in weight-bearing position and others are usually taken with non-weight bearing position. In conclusion, despite some disadvantages including being uncomparable to other measurements, including several joint movements, and untested validity, this method should not be ignored when relationships of this measurement are tested against other variables taken in functional tasks, such as running, jumping, and landing.

Jump-landing tasks

Two types of landing tasks were tested; drop-jump landing (DJL) and forward-jump landing (FJL). Participants were allowed to practice each jump-landing task until they felt comfortable to perform. However, no specific instruction was provided regarding the landing tasks to minimize the coaching effect on the participant’s natural landing strategies. Minimum of one complete jump-landing sequence was verified by the co-investigator before testing trials to make sure understanding of the each task.

Verbal instruction for landing tasks

The co-investigator verbally explained the sequence of the landing tasks, but no specific instruction was provided regarding how to land in order to observe participant's natural landing mechanics. Detailed instruction of the sequence could possibly alter participant's landing mechanics because they realized what was being examined. The co-investigator emphasized four key characteristics in the tasks. First, for DJL, the participant dropped off the box without jumping up vertically, whereas for FJL, the participant jumped off forward without jumping up. Hence, DJL task was performed as a relatively vertical movement that simulates a landing from a rebounding in basketball. On the other hand, FJL task was performed as a relatively horizontal movement that simulates landing during a jump-stop in basketball. Second, the participant landed straight in front of the box facing the front digital camera to be recorded appropriately. Third, the participant vertically jumped up as high as they could immediately after they landed from the box. Fourth, the participant performed the tasks in a smooth sequence.

Drop-jump landing task

The Drop-jump landing (DJL) protocol followed the previously used protocol[47] with some modifications. The five-step DJL sequence was as follows; (1) participant extended the right leg in forward with approximately 30° of hip flexion, (2) participant took off with the left foot without jumping up movement, (3) participant landed on the force platform immediately below the extended right foot; hence the task was relatively vertical landing, (4) participant performed a maximum vertical jump immediately after two-foot landing, and (5) participant landed back on the platform.

Forward-jump landing task

The Forward-jump landing (FJL) protocol followed the previously used protocol[46] with a modification. The five-step FJL sequence was as follows; (1) participant jumped off from the box with two-feet, (2) participants jumped forward, but not vertically, to reach the center of the force platform, (3) participant landed on the center of the platform with both feet, (4) participants performed a maximum vertical jump immediately after two-foot landing, and (5) participants landed back on the platform.

Validity and reliability of two-dimension kinematic assessment

Many of previous studies that investigated landing kinematics and kinetics used a three-dimensional motion analyze system synchronizing with a force plate. Although the three-dimensional motion analyze system is expensive, complex, and time-consuming, the system has been considered as a "gold standard" and used to evaluate human movement.[45,152,153,165,166] Commercially available digital camera has also been used to assess jump-landing technique and lower extremity injury risk screening in clinical settings where three-dimensional movement evaluation was not feasible.[2,46,47,146,149,152,153,166,167] Using commercially available digital cameras might be relatively easy for evaluators regardless of experience to select an appropriate frame, but researchers pointed out that precisely filming human kinetics, especially high-speed movement, was difficult by using commercially available digital cameras due to low frame rate (30 to 60 Hz).[152,168]

In addition to the difficulty of capturing precise human movement, several investigators indicated that commercially available video cameras were unable to detect transvers plane kinematics even though two video cameras were used to record single-plane frontal and/or sagittal plane motions.[47,140,153,166] Nevertheless, several authors claimed that apparent frontal plane joint angles were affected by the hip, knee, and ankle joints' sagittal, frontal, and transverse plane movements including hip flexion, femoral internal rotation, knee valgus, knee anterior translation, tibial external rotation, ankle flexion, and foot rotation.[45,47,140,153,166] Therefore, because apparent frontal plane knee angle does not represent true knee valgus or abduction, the observed frontal plane knee angle has been referred as various operational terms, such as medial knee collapse[140], frontal plane knee projection angle (FPPA)[166], dynamic knee valgus[153,166], or medial knee displacement (MKD).[169]

An alternative to gold standard three-dimensional motion analysis system is the two-dimensional video analysis. Even though the single-plane two-dimensional analysis were usually unable to synchronize to data from a force plate or other commercially available video recorder, the use of two video recorders could provide frontal and sagittal plane kinematic information for landing mechanics evaluation. The two-dimensional video digitizing was completed by using computer software that required digitization of selected frame of captured video manually.[45,130]

When single-plane video recorded clips was evaluated by a two-dimensional digitizing software, sagittal plane kinematics were valid against universal goniometric measurements, but frontal plane kinematic compared to three-dimensional motion analysis were inconclusive. Concurrent validities of two-dimensional assessment of sagittal plane hip and knee joint angles compared to goniometric measurements were excellent (hip flexion: $r = .95$; knee flexion: $r = .98$).[165] In this study, the sagittal plane joint angles were assessed during an isometric mechanical lifting task measured with a back leg chest dynamometer system. The hip angle was formed by two lines; one line was running between the lateral aspect of the acromion process and the greater trochanter, the other was a line between the greater trochanter and the lateral femoral epicondyle. The knee angle was formed by two lines; one was running between the greater trochanter and the lateral femoral epicondyle, and the other was connecting the lateral femoral epicondyle to the lateral malleolus.[165]

Another study that investigated validity of sagittal plane lower extremity joint angles compared to goniometric measures also reported that two-dimensional kinematic analysis was moderately valid.[130] The ankle and knee sagittal plane angles during a functional movement, Star Excursion Balance Test, showed less than 4° measurement error, and the hip flexion angle measurement error was less than 11°.[130] Nonetheless, the validity of two-dimensional kinematic assessment has not been compared to the gold-standard, three-dimensional motion analysis to date. To validate the sagittal plane kinematic analysis with two-dimensional video assessment, the results need to be compared to the results of three-dimensional motion analysis.

The sagittal plane lower extremity kinematics also showed moderate to high intra-rater, inter-rater, and test-retest reliabilities during functional tasks. During an isometric mechanical lifting task, the intra-rater reliability of hip flexion was high ($ICC_{3,1}$ = 0.99), and that of knee flexion was high ($ICC_{3,1}$ = 0.98), as well. Inter-rater reliabilities of hip flexion and knee flexion were also high ($ICC_{2,1}$ = 0.96, and $ICC_{2,1}$ = 0.96, respectively).[165] During Star Excursion Balance Test, the test-retest reliability of the peak ankle, knee, and hip angles were high ($ICC_{3,1}$ = 0.82, $ICC_{3,1}$ = 0.85, and $ICC_{3,1}$ = 0.79, respectively).[130]

While the validity of sagittal plane kinematics was compared to universal goniometric measurements, validity of frontal plane kinematics was compared to three-dimensional motion assessment. Moderate to high correlations between two-dimensional kinematic analysis and three-dimensional kinematic analysis have been reported. Concurrent validity of frontal plane knee angle measured by two-dimensional video digitizing evaluation demonstrated high against three-dimensional motion analysis during side step (r = 0.76) and side jump (r = 0.80) tasks.[45] Pearson correlation coefficient of FPPA compared to knee external rotation and hip adduction during a single-leg squat was moderate with r = 0.48 and was poor with r = 0.32, respectively. In addition, FPPA was also associated with hip adduction (r = 0.61), and knee abduction (r = 0.49).[170] On the other hand, low concurrent validity of two-dimensional assessment at knee and hip joints were also reported against three-dimensional motion analysis. Compared to FPPA, poor Pearson correlation coefficients were demonstrated in knee flexion (r = -0.26), knee abduction (r = -0.20), knee internal rotation (r = 0.12), hip flexion (r = -0.32), hip abduction (r = 0.34), and hip internal rotation (r = 0.15).[169] Thus, a two-dimensional joint kinematic assessment might be an alternative to measure knee joint medial displacement during functional tasks. It must be noted that transverse plane joint rotation could not be identified by two-dimension assessment.

Intra-rater reliability of two-dimensional video digitizing was reported that simple task such as countermovement vertical jump demonstrated better reliability than more complex tasks such as drop vertical jump task. The intra-rater reliability of FPPA measured by two-dimensional digitizing method was high in functional tasks including two-leg countermovement vertical jumps (r = 0.75), single-leg countermovement jumps (r = 0.64), single-leg side spring (r = 0.75), and single-leg drop vertical jump (r = 0.54). The intra-rater reliability in measuring the angle between the line of femur and horizontal line was also high in the same tasks; two-leg countermovement vertical jumps (r = 0.74), single-leg countermovement jumps (r = 0.68), single-leg side spring (r = 0.73), and single-leg drop vertical jump (r = 0.60).[171] It was concluded that simplicity of functional tasks could increase in intra-rater reliability[171] possibly due to insufficient recording frame rate of video recorders that was usually between 30 and 60 Hz.

In addition to simplicity of tasks, easiness to identify anatomical landmarks during manual digitizing process could contribute to the degree of reliability in two-dimensional kinematic analysis. Reliability of two-dimensional manual video digitizing with and without using 2cm-diameter marker was compared in treadmill running, and the result reported that joints that was visually exposed including shoulder (marker (M) $\omega_2 = 0.82$; non-marker (NM) $\omega_2 = 0.77$), elbow (M $\omega_2 = 0.76$; NM $\omega_2 = 0.68$), and knee joints (M $\omega_2 = 0.97$; NM $\omega_2 = 0.67$) showed higher reliability than joints that was covered or hidden such as hip joints by shorts (M $\omega_2 = 0.92$; NM $\omega_2 = 0.41$) and ankle joints by shoes (M $\omega_2 = 0.70$; NM $\omega_2 = 0.16$), where ω_2 was contribution of intra-tester variability to total variability.[172] Hence, without using marker on selected anatomical landmarks, joint kinematics could not be reliably measured, especially the joints that were not visually seen.

In conclusion, apparent frontal plane knee displacement could be validly and reliably identified with two-dimensional motion evaluation, but accurate knee valgus angle that is largely affected by hip and knee joint kinematics should not be measured by two-dimensional video digitizing method. Using high-speed camera that is able to capture over 200 Hz may accurately select anatomical landmarks during two-dimensional video digitizing; therefore, high frame rate could improve accuracy of single-plane motion analysis. Also, filming dynamic movements with two cameras could capture separate frontal and sagittal plane joint kinematics, even though manual time synchronization of two videos and with force plate data is not feasible without a sophisticated digitizing computer system.

Isokinetic strength measurement

Following the landing tasks, isokinetic strength of plantar flexors in both ankles will be measured. The position of a participant was seated on the tilted dynamometer chair (85° reclining) with the hip joint flexed approximately 90° flexion, and the distal femur was supported on a supplemental pad by using an appropriate strap. The subject's knee joint was maintained at 70° flexion during the testing. The axis of the dynamometer was aligned to the body of the talus, lateral malleolus, and just below the medial malleolus so that the anatomical rotational axis of the ankle was aligned with the mechanical rotational axis of the dynamometer. After adjusting the height of the dynamometer and the distance of the seat from the dynamometer, the foot was strapped on a footplate which was connected to the axis of the dynamometer.

Isokinetic torque of the plantar flexors was recorded between 30° of plantar flexion and 10° of dorsiflexion from the anatomically neutral position of the ankle. Concentric torques were measured at 60°/s, 120°/s, and 180°/s, while isokinetic eccentric torque was recorded at 30°/s. For each angular velocity, up to five repetitions of sub-maximal practice trial were allowed prior to the maximal test trial, and the isokinetic testing consisted of three repetitions with maximal effort. Each testing was followed by two minutes of recovery period. During the isokinetic testing, constant verbal encouragement ("push down the foot as hard/ fast as you can") was provided to help participants produce maximum effort during the testing. For each angular velocity, the peak torque value was analyzed.

Isokinetic dynamometer allows objectively measure muscular functional variables by generating a counter-torque equal to the force produced by a subject.[173] The measurable variables include, but not limited to, peak torque, average peak torque, angle-specific torque, power, and work by generating a constant velocity throughout a pre-determined ROM. It was reported due to constant movement velocity, isokinetic muscular function testing was reliable and valid when the angular velocities was less than moderately high (< 300°/s).[174,175]

Data Collection and Analyses

Data collection

Participants were allowed to practice the tasks until they could perform the sequence properly. Minimum of one complete sequence was verified by the co-investigator prior to each task. Three ankle brace conditions (no brace = NB, LTB, and HTB) were counterbalanced for all participants. Two landing tasks (DJL and FJL) were also be alternated for all participants. Two measurements were taken for each ankle condition and task, and the measurements were averaged for analyses. Therefore, each participant completed a total of 12 landing tasks.

The digital camera captured the landing tasks at a sampling rate of 210 Hz. The kinematic data were stored in a memory cards mounted in each digital camera. The data stored in memory cards were transferred to a laptop computer for analyses using Dartfish Motion Analysis Software (version 10.0; Dargfish®, Fribourg, Switzerland). Force plate data were collected at a sampling rate of 400 Hz. The force platform was connected to a laptop computer so that the GRF data were stored in the computer simultaneously for analyses by using computer software (AccuPower version 1.5; Athletic Republic, Fargo, ND).

Data analysis

Three dependent variables were mainly analyzed in this research project. Medial knee displacement was operationally defined as the ratio of distance between the center of left and right patellae relative to the distance between the left and right anterior superior iliac spines. In other words, the distance between the patellae was normalized to the distance between the anterior superior iliac spines. The distance of those landmarks are generally dependent on participant's body size; therefore, the ratio, instead of actual distance, was considered as better measurement. This analysis was performed on a computer screen with Dartfish Video Motion Analysis software. Values from the two measurements in each conditions and tasks were averaged for statistical analyses.

Sagittal plane ankle angle was measured only on the right side of the body at two moments during an initial landing task from the take-off box. The first was at the initial foot contact where any part of the foot made contact with the surface of the force platform. The second was measured at maximum dorsiflexion of the ankle. The total angle between the initial foot contact and the maximum dorsiflexion during a landing task was defined as ankle joint displacement. The optimal references were the shaft of the lower extremity and the shaft of the 5^{th} metatarsal. The first line was the shaft of the lower extremity that was referenced between the center of the knee joint line and the lateral malleolus. Because the shoe covered the shaft of the 5^{th} metatarsal, the landmark was assumed to be parallel to the sole of the shoe. Therefore, the second line was drawn from the lateral malleolus toward the toes that was parallel to the sole of the shoe. Values from the two trials in each conditions and tasks were averaged for statistical analyses.

Maximum ground reaction force (GRF) was obtained during a landing task. AccuPower software was used to analyze the GRF. Raw GRF value was re-calculated relative to each participant's body weight (in Newton). Because participant's GRF is dependent on their body weight, the force value needed to be normalized to the body weight. To maintain the landing height relatively constant, the co-investigator emphasized not to jump up as participants took off the box. Values from the two trials in each conditions and tasks were averaged for statistical analyses.

Isokinetic data collected by the dynamometer were analyzed by Biodex Advantage software (version 4.47; Biodex Medical System, Shirley, NY). Isokinetic strength was determined by peak torque, and the peak torque was defined as the maximum torque recorded in the five testing repetitions and expressed in Newton-meters (N·m).[176] The peak torque was, then, normalized to the body weight to evaluate a relationship between the isokinetic torque and GRF. When no side difference is observed, the left and right peak torques were combined and averaged because the landing tasks were performed by both feet and because the GRF value was expressed as a two-feet landing force.

Statistical Analyses

The sample size was calculated based on the pilot study in which the minimum expected mean difference and *SD* in peak ankle dorsiflexion during a DJL task between non-braced (NB) condition and high-tensioned braced (HTB) condition are 3.47° and 3.9°, respectively. Based on this estimation, the total number required was 20 subjects in each condition. The equation used to calculate the required sample size was;

$$N = \frac{4\sigma^2(z_{crit} + z_{pwr})^2}{D^2}$$

The means and *SDs* for each dependent variable were calculated across the two trials for each condition and task. The days after the menstrual period were also reported. Prior to evaluating the hypotheses, $ICC_{3,2}$ were be calculated to examine the reliability of measurements. For the (1) and (2) hypotheses, the effect of EAS on ankle displacement, tMKD, and vGRF were evaluated by performing a repeated-measures MANOVA because these dependent variables were believed to be associated. Multivariate normality was tested by evaluating univariate normality for each dependent variable. Homogeneity of covariance was checked by Levene's test. If omnibus *F* was significant, separate ANOVAs on each dependent variable with Tukey post-hoc test was performed. Following the assumption checking procedures, interaction was evaluated. When interaction was not significant, separate repeated-measure ANOVAs with Tukey multiple comparison post-hoc test. For the (3) and (4) hypotheses, the effect of EAS and landing tasks on ankle displacement, knee displacement, peak MKD, vGRF, and postGRF were evaluated by performing a repeated-measure 3 (ankle brace conditions) x 2 (landing tasks) MANOVA. Following the assumption checking procedures, interaction was evaluated. When interaction was not significant, separate repeated-measure ANOVAs with post-hoc Tukey multiple comparison test at each level were conducted to examine the main effects. Following conducting necessary ANOVA, η^2 of effect size was calculated to assist clinical interpretation of the independent variables on the dependent variables. The effect size η^2 was defined as follows; trivial < 0.01, small < 0.06, medium < 0.14, and $0.14 \leq$ large.[177]

Additionally, Pearson's correlation coefficients (r) were computed to evaluate relationships among ankle ROM and ankle displacement, among the ankle ROM, ankle displacement, knee displacement, hip displacement, tMKD, peak MKD, vGRF, postGRF, and plantar flexor isokinetic peak torque. The α level is set a priori at $\leq .05$ for all statistical analyses. Statistical Package for the Social Sciences (version 20.0, SPSS Inc, Chicago, IL) will be used to analyze the data.

PAPER ONE: ANKLE BRACING INCREASES MEDIAL KNEE DISPLACEMENT AND VERTICAL GROUND REACTION FORCE DURING A DROP-LANDING TASK

Abstract

Context: Anterior cruciate ligament (ACL) injuries can occur during landing or deceleration tasks. Sagittal plane ankle range of motion (ROM) appears not only to play important role in absorbing ground reaction force (GRF) but also to affect medial knee displacement (MKD) during the maneuvers. In addition to the frontal plane ankle ROM, external ankle supports (EAS) can restrict the sagittal plane ankle ROM, but how EAS affects the landing mechanics is not well understood. **Objective:** To evaluate the effects of EAS on landing kinematics and kinetics. **Design:** Crossover study. **Setting:** Controlled laboratory environment. **Patients or Other Participants:** Nineteen physically active females [*M* (*SD*): age = 20.2 (1.1) years, height = 170.0 (7.15) cm, mass = 65.7 (8.0) kg]. **Intervention(s):** Participants performed a drop-landing task under three bracing conditions: no bracing, low-tensioned bracing, and high-tensioned bracing. **Main outcome Measure(s):** Static knee extended and knee flexed ankle ROMs, sagittal plane ankle displacements, total MKD (tMKD), vertical GRF (vGRF), plantar flexor isokinetic peak torque (concentric contraction: 60°/s, 120°/s, and 180°/s, and eccentric contraction: 30°/s). **Results:** Ankle bracing conditions significantly altered ankle displacement ($F_{2,36} = 15.42$, $P < .001$), tMKD ($F_{2,36} = 12.56$, $P < .001$), and vGRF ($F_{1.195,21.515} = 5.72$, $P = .021$). However, planter flexor isokinetic peak torque demonstrated no correlation with vGRF or tMKD regardless of the angular velocities. **Conclusions:** Excessively restricted sagittal plane ankle ROM by ankle bracing reduced the ankle displacement, and increased frontal plane knee displacement and vGRF. Therefore, healthcare professionals should be aware of not excessively limit the sagittal plane ankle ROM as they apply ankle taping or bracing.

Introduction

Knee ligamentous injuries, including the anterior cruciate ligament (ACL) rupture were more common among female athletes, especially in basketball and soccer.[1] Two major tasks performing during non-contact ACL tear were cutting and landing tasks.[35] It is believed that landing with increased knee valgus angle and greater ground reaction force (GRF) produces greater ACL strain during a deceleration task.[37] Therefore, greater knee valgus angle and greater GRF could lead to non-contact ACL injury because they appeared to be associated with subsequent loading of the ACL during landing or deceleration tasks.

Compared to non-ACL injured subjects, ACL injured subjects demonstrated less plantar flexion of the ankle at initial contact phase of landing[35]; in other words, they showed heel first landing or flat foot landing.[37] Non-contact ACL injuries are also common in skiing, and 60% of ACL injury occurred when the skier was attempting to change direction, and 85% of ACL injury occurred when the ski boot binding was not released so the impact from the ground was not absorbed at the ankle joint.[29] This high occurrence attributed to the structure of the rigid ski boots that does not allow the ankle motion. Hence, the ankle joint appears to play an important role in decelerating and changing a direction to absorb impact from a playing surface.

Ankle range of motion (ROM), particularly dorsiflexion, affects the loading stress on the knee joint. The GRF forced the ankle joint into dorsiflexion, and the eccentrically contracting plantar flexor muscles counteracted the dorsiflexion to absorb GRF during a toe landing task.[108] When a toe landing was compared to a heel landing, toe landing strategy showed greater ankle displacement and less peak GRF than heel-landing technique.[7] Thus, landing with less ankle displacement could be associated with the greater risk of non-contact ACL injury due to lack of capacity to attenuate landing GRF. Greater sagittal plane joint ROM appeared to reduce vertical

GRF (vGRF) during landing tasks. The lower extremity muscles played important roles to absorb landing force,[178] and angular displacement at the ankle joint and strength of the plantar flexors were inversely associated with vGRF.[3] Therefore, it was speculated that greater plantar flexion and subsequent greater ankle displacement maximally utilized the plantar flexion muscle to absorb vertical landing force.

In addition to abnormally greater GRF, one of the non-contact ACL injury mechanisms is the excessive valgus angle at the knee joint.[2] The commonly observed mechanism of non-contact ACL injury was “valgus collapse” of the knee joint. The valgus collapse is a combination of knee abduction, external rotation of the tibia, and internal rotation of the femur.[179] This frontal plane knee kinematics appeared to be affected by the sagittal plane joint kinematics.[39] Medial knee displacement (MKD) was operationally defined as apparent medial movement of the knee joint in the frontal plane which involved the hip, knee, and ankle joints’ sagittal, frontal, and transverse plane movements.[47] Although 2-dimension analysis can not identify rotational movement, it can be used to distinguish excessive knee valgus, a movement in the frontal plane.[45] In addition to MKD, GRF could be increased by smaller ankle ROM and smaller plantar flexor strength during a landing task.[4] No clear explanation has been made which plantar flexor muscle was mainly associated to the increased MKD, but plantar flexion strength might more rely on the soleus strength as the knee flexion angle increases.[89] One possible explanation was that the gastrocnemius loses the plantar-flexing moment as the knee joint increase its angle because it is a two-joint muscle crossing the ankle joint and the knee joint.[3] Therefore, the gastrocnemius might not be fully used to attenuate a landing force during natural landing due to knee flexion.

The external ankle supports (EAS), including ankle taping and bracing, have been widely used to prevent ankle sprain in various physical activities. Although the main purpose of EAS is to limit the frontal plane ankle ROM to reduce the risk of inversion or eversion ankle sprain, the application of EAS also restricts the sagittal plane ankle motions.[41] This restricted sagittal plane ankle ROM could alter the landing kinematics and kinetics and could consequently increase the risk of non-contact ACL injury. Smaller dorsiflexion during a deceleration phase of activities appeared to increase subsequent GRF[3,7] and MKD,[39] and this could increase the risk of an ACL injury. The influence of the ankle ROM limited by an external support in terms of the dynamic knee valgus during a landing, however, has not been studied to date. Restricting the ankle displacement could also alter the landing mechanics. Therefore, the purpose of this study was to investigate whether the limited ankle ROM by application of EAS changes landing kinematics and kinetics during a drop-jump landing. Two hypotheses were investigated: first, limiting ankle ROM using external ankle supports (EAS) would decrease ankle displacement, increase total MKD (tMKD), and increase vGRF during a drop-landing task. Second, greater isokinetic plantar flexor strength would be inversely correlated with tMKD and vGRF.

Methods

Participants

Nineteen physically active and healthy college aged female subjects [*M* (*SD*): height = 170.0 (7.15) cm; mass = 65.7 (8.0) kg, age = 20.2 (1.1) years, 11.4 (9.2) days post menstrual period] without history of lower extremity injury completed the test protocol. The inclusion criteria were participation in physical activity at least twice per week and minimum of 30 minutes per session, no current neuromuscular dysfunction, no signs or symptom of inflammation, no history of surgeries within six months in the lower extremities. The participants were recruited from a local university and completed a demographic form and signed an informed consent prior to being tested. The study was approved by the university's Institutional Review Board.

Ankle Range of Motion Measurement

After completing warm-up exercises, ankle ROM was measured in non-weight bearing and weight bearing positions. The non-weight bearing active ROM of the left and right ankle was measured in two positions. Knee extended ankle ROM was measured by the subject sitting on a table and the knee joint kept at full extension. Knee flexed ankle ROM was measured by the subject sitting with flexing the hip and knee at 90°. Two measurements were taken in each position by the same researcher and averaged for analyses. In weight bearing position, lunge dorsiflexion was measured by determining the angle of the lower leg relative to the vertical line (Figure 1).[158] Because the torque applied to the ankle joint at the weight bearing position was greater than that applied by an examiner's hand, the advantage of this method was that the measurement could more represent the ankle dorsiflexion during functional tasks, such as running, jumping, and landing.[158]

External Ankle Supports Application

Prophylactic lace-up ankle braces (ASO® Ankle Stabilizer; Medical Specialties, Inc., Charlotte, NC) were fit on the subjects' ankles as instructed by the manufacturer. During the brace application, participants were sitting in a chair and the lower leg was held perpendicular to the floor so that the ankle joint was maintained in 90°. Each lace was tied with a relatively constant tension at 2.5 – 3.0 kg (low tension brace, LTB) or 7 – 7.5 kg (high tension brace, HTB) measured by a hand-held digital scale (Rapala VMC, Vaaksy, Finland). Each lace was held between the target tensions for one second at 90° angle from the surface contour of the ankle braces.

Drop-Landing Task

Participants were instructed to wear athletic shirt, shorts, and low-cut sport shoes. For frontal plane kinematic recordings, markers were placed at both left and right side of the body at 2 cm anterior to the acromioclavicular joints, at the anterior superior iliac spine, at the center of patella, and at the middle of the distal lower extremity 5cm superior to the ankle joint. For sagittal plane kinematic recordings, markers were placed only on the right side of the body at 2 cm lateral to the acromioclavicular joint, at the greater trochanter, at the center of the knee joint line, and at the lateral malleolus. Two digital cameras (EX-FH20; Casio, Inc., Tokyo, Japan) were used to record landing kinematics. Digital cameras were mounted on a tripod 73 cm from the floor. The tripod was placed 175 cm in front of the center of the force plate. This digital camera recorded tMKD in frontal plane. Another tripod with a digital camera mounted on was positioned at 175 cm on the right side of the center of the force plate. This second digital camera recorded ankle and knee displacements in sagittal plane. The force platform (AccuPower; AMTI, Watertown, MA) was embedded to the testing surface.

The drop-jump landing protocol followed Noyes's protocol[47] with some modifications. A 33cm box was placed 5cm from the back edge of the platform, which was 35 cm from the landing target on the platform. The five step drop-jump landing sequence was as follows: (1) participant extended the right leg in forward motion with approximately 30° of hip flexion; (2) participant took off with the left foot without jumping up movement; (3) participant landed with both feet simultaneously on the force platform with facing the front digital camera to be recorded properly; (4) participant performed a maximum vertical jump immediately after two foot landing; and (5) participant landed back on the platform. Prior to a trial jump, participants were allowed to practice the jump-landing task and provided two-minute rest. They practiced the task until they felt comfortable to perform it without hesitation. The sequence of the landing task was explained; however, no specific instruction was provided regarding the landing tasks to minimize the coaching effect on the participant's natural landing strategies.

Isokinetic Strength Measurement

Following the landing tasks, isokinetic strength of plantar flexors in both ankles was measured. The Biodex Multi-Joint system 4 PRO (Biodex Medical System, Shirley, NY) was used to assess concentric and eccentric isokinetic torque of the plantar flexors. The position of a participant was seated on the tilted dynamometer chair with the hip joint flexed at 90° flexion, and the distal femur was supported on a supplemental pad by using an appropriate strap. The participant's knee joint was maintained at 70° flexion during the testing. The axis of the dynamometer was aligned to the body of the talus, lateral malleolus, and just below the medial malleolus so that the anatomical rotational axis of the ankle was aligned with the mechanical rotational axis of the dynamometer. After adjusting the height of the dynamometer and the distance of the seat from the dynamometer, the foot was strapped on a footplate which was connected to the axis of the dynamometer.

Isokinetic torque of the plantar flexors was recorded between 30° of plantar flexion and 10° of dorsiflexion from the anatomically neutral position of the ankle. Concentric torque was measured at 60°/s, 120°/s, and 180°/s, while isokinetic eccentric torque was recorded at 30°/s. For each angular velocity, five repetitions of submaximal practice trial were allowed prior to the maximal test trial. The isokinetic testing consisted of three repetitions with maximal effort. Each testing angular velocity was followed by two minutes of recovery period. During the isokinetic testing, constant verbal encouragement was provided to help participants produce maximum torque during the testing. For each angular velocity, the peak torque value was analyzed.

Data Collection

Three ankle brace conditions (no prophylactic ankle bracing = NB, low tension ankle bracing = LTB, and high tension ankle bracing = HTB) were counterbalanced for all participants. The digital cameras captured the landing tasks at a sampling rate of 210 Hz. The kinematic data stored in memory cards were transferred from each camera to a laptop computer for kinematic analyses using Dartfish Motion Analysis Software (version 10.0; Dargfish®, Fribourg, Switzerland). Force plate data were collected at a sampling rate of 400 Hz. The force platform was connected to a laptop computer so that the GRF data were stored in the computer for kinetic analyses by using computer software (AccuPower version 1.5; Athletic Republic, Fargo, ND).

Data Analysis

<u>Kinematic measurements</u>

The tMKD represents the total medial displacement of the patellae (the distance of the patellae at the initial foot contact subtracted from the distance of the patellae at the maximum knee flexion; Figure 2). The distance of the landmarks was generally dependent on participant's body size; therefore, the normalized ratio relative to the anterior superior iliac spines was considered as better measurement than the actual distance between the patellae. Values from the two measurements in each condition were averaged for statistical analyses.

Ankle displacement was defined as the total angle between the foot contact and the maximum dorsiflexion during a landing task. The sagittal plane ankle displacement was measured only on the right side of the body at two instants during an initial landing from the

33cm box. The first moment was at the foot contact when any part of the right foot made contact with the surface of the force platform. The second moment was measured at maximum dorsiflexion of the ankle. Total angle between the two instants was calculated as ankle displacement. Ideally, the references for ankle angle measurement were the shaft of the lower extremity and the shaft of the 5^{th} metatarsal. The first line was the shaft of the lower extremity that is referenced between the center of the knee joint line and the lateral malleolus. Because the 5^{th} metatarsal was covered by the shoe, the landmark was assumed to be parallel to the sole of the shoe; therefore, the second line was drawn from the lateral malleolus toward the toes that is parallel to the sole of the shoe. Values from the two trials in each condition were averaged for statistical analyses.

Kinetic measurements

Peak vertical GRF (vGRF) was obtained during the landing task. AccuPower software was used to analyze the vGRF. Raw vGRF value was re-calculated relative to each participant's body weight (BW in N). Values from the two trials in each condition were averaged for statistical analyses. Isokinetic data collected by the dynamometer were analyzed by Biodex Advantage software (version 4.47; Biodex Medical System, Shirley, NY). Isokinetic peak torque was normalized to the body mass (BW in kg) to evaluate a relationship between the isokinetic peak torque (N/kg) and vGRF (N/BW).[176] Because no side difference was observed, the left and right peak torque were combined and averaged, and the GRF value was expressed as a two-feet landing force.

Statistical Analysis

The sample size was determined based on the pilot studies in which the minimum expected difference and *SD* in ankle dorsiflexion during a drop-jump landing task between NB condition and HTB condition were 3.47° and 3.9°, respectively. The means and *SD* for each dependent variable were calculated across the two trials for each condition. The effect of EAS conditions on ankle displacement, tMKD, and vGRF were evaluated by performing a repeated measure MANOVA. Following the assumption checking procedures, interaction was evaluated. When interaction was not significant, separate repeated measures ANOVAs with Tukey multiple comparison post hoc test were performed. Following conducting necessary ANOVAs, η^2 of effect size was calculated to assist clinical interpretation. The effect size η^2 was defined as follows; trivial < 0.01, small < 0.06, medium < 0.14, and 0.14 ≤ large.[177] Pearson's correlation coefficients (*r*) were also computed to evaluate relationship between the plantar flexor isokinetic peak torque and vGRF, and between the plantar flexor isokinetic peak torque and tMKD. The α level was set a priori at ≤ .05 for all statistical analyses. Statistical Package for the Social Sciences (version 20.0, SPSS Inc, Chicago, IL) was used to analyze the data.

Results

Two hypotheses were investigated in this study: first, limiting ankle ROM using EAS would decrease ankle displacement, increase tMKD, and increase vGRF during a drop-landing task; second, greater isokinetic plantar flexor strength would be inversely correlated with tMKD and vGRF. A MANOVA was conducted to assess if there were differences between the three ankle brace conditions (NB, LTB, and HTB) on ankle displacement, tMKD, and vGRF during the drop-jump landing task. The assumptions of MANOVA were checked and met. A significant multivariate effects were found in ankle brace conditions (Wilks' $\Lambda = .725$, $F_{4,8} = 2.23$, $P = .03$, observed power = .84). Because omnibus F value of MANOVA was significant, we further conducted separate repeated-measure ANOVA on each dependent variable.

Follow-up repeated measures ANOVAs with Tukey multiple comparison post hoc test was conducted to assess whether there were differences in ankle displacement, tMKD, and vGRF among the ankle brace conditions. Results showed that ankle displacement, tMKD, and vGRF were significantly different for ankle brace conditions. The ankle displacement during the drop-jump landing task was significantly different among ankle brace conditions ($F_{2,36} = 15.42$, $P < .001$, observed power > .99, $\eta^2 = .22$). The tMKD during the landing task was different among ankle brace conditions ($F_{2,36} = 12.56$, $P < .001$, observed power = .99, $\eta^2 = .08$). The results of a repeated measures ANOVA with Greenhouse-Geisser adjustment indicated vGRF during the landing task was significantly different among ankle brace conditions ($F_{1.195,21.515} = 5.72$, $P = .021$, observed power = .67, $\eta^2 = .02$). The descriptive statistics and Tukey post-hoc multiple comparison test results of each dependent variable are shown in Figure 3 and Table 1, respectively. Hence, our findings supported the first hypothesis of this study.

To evaluate the second hypothesis whether plantar flexor isokinetic peak torque (concentric contraction: 60°/s, 120°/s, and 180°/s, and eccentric contraction: 30°/s) was correlated with vGRF and tMKD, Pearson's correlations (r) were computed. Results indicated plantar flexor isokinetic peak torque did not demonstrated a correlation with vGRF (60°/s: $r = .10$, $P = .10$, 120°/s: $r = .002$, $P = .99$, 180°/s: $r = .05$, $P = .83$, -30°/s: $r = .08$, $P = .74$) and with tMKD (60°/s: $r = .11$, $P = .65$, 120°/s: $r = .20$, $P = .41$, 180°/s: $r = .17$, $P = .48$, -30°/s: $r = -.08$, $P = .74$). Thus, the second hypothesis was rejected.

Discussion

The purpose of this study was to investigate whether the limited ankle ROM by application of EAS changes landing kinematics and kinetics. Although external ankle supports (EAS) are used to limit the frontal plane ankle ROM to reduce the risk of ankle sprain, the application of EAS also restricts the sagittal plane ankle motions.[8,41] This restricted sagittal plane ankle ROM appeared to alter the landing kinematics and kinetics.[5] Smaller dorsiflexion during a deceleration phase of activities was found to increase subsequent MKD,[4,141] and this limited dorsiflexion in a landing task could increase the risk of non-contact ACL injury due to increasing MKD or knee valgus.

In this study, the laces of the ankle braces were controlled in two tensions. The laces were tied with 3 -5 kg in LTB condition. A majority of the subjects (17/19 = 89%) commented that LTB condition was felt “snug.” On the other hand, in HTB condition, the laces were tied with 7 – 8 kg. Only a small portion of the subjects (2/19 = 11%) preferred HTB conditions. The effect of EAS to restrict the sagittal plane ankle displacement during drop-jump landing was consistent with the results of a previous study[134] that also investigated the effect of ASO prophylactic ankle brace®. The study reported that the means of ankle joint displacement during a jump-landing task were 59.3 (14.3)° and 56.5 (12.5)° without and with EAS, respectively.

Our results indicated that EAS application reduced the ankle displacement, increased tMKD, and increased vGRF during a drop-landing task; thus, the first hypothesis was supported. In contrast, the second hypothesis was not supported. We found that isokinetic plantar flexor strength (concentric contraction: 60°/s, 120°/s, and 180°/s, and eccentric contraction: 30°/s) was not correlated with tMKD or vGRF. The primary finding in this study was EAS application increased tMKD during a drop-landing task. Bell and colleagues[4] found that subjects with smaller knee-flexed dorsiflexion ROM exhibited greater MKD in which the patella moved medially to the great toe during a two-leg squat. They compared a group of greater MKD and a group of less MKD. Greater MKD group showed the excessive medial knee movement which the authors defined as the patella moving medially to the great toe during a two-leg squat. The researchers found that greater MKD group had smaller dorsiflexion ROM than control group. They also found that the greater MKD group reduced MKD when performing a two-leg squat with a 5.1cm heel lift under the calcaneus. The use of the artificial heel lift increased the plantar flexion ROM before the squatting exercise, and the angle of the lower leg was altered which was similar to increased dorsiflexion.[4] Macrum et al.[141] also reported that restricted ankle dorsiflexion increased MKD compared to control group. They compared MKD during a two-leg squat in two conditions; squatting on a 12° posteriorly slanted board and no slant board. With posteriorly slanted board, the ankle joint was put into dorsiflexed position; in other words, the subjects started the squat movement at dorsiflexed position. Consequently, the dorsiflexion displacement was reduced during the squat task. As a result, subjects with limited dorsiflexion decreased knee flexion and increased MKD.[141] Bell et al.[4] and Macrum et al.[141] measured MKD during a controlled squat movement. In contrast, we measured tMKD during a dynamic movement and still found the result was consistent with the previous studies.[4,141] Artificially limited ankle available ROM in both plantar flexion and dorsiflexion resulted in reduction of ankle displacement during the dynamic task. We speculated that limited peak ankle dorsiflexion [NB: *M* (*SD*) = 26.7 (4.6)°, LTB: *M* (*SD*) = 23.1 (4.5)°, HTB: *M* (*SD*) = 21.8 (5.0)°] increased tMKD during a drop-landing task. It was proposed that reduced dorsiflexion prevented the body from lowering the center of mass during a weight-bearing activity, which induced greater pronation at the subtalar joint and subsequent greater internal rotation of the tibia and knee valgus angle.[180] In a cadaveric simulation study, landing with increased knee valgus angle induced greater ACL strain compared to a landing without knee valgus.[120] Our results indicated that increased tMKD by EAS may increase a knee valgus loading and ACL strain during a drop-landing task.

We found that limited ankle ROM also decreased the ankle dorsiflexion during the drop-landing task. Mcrum et al.[141] discovered that limited dorsiflexion with a posteriorly slanted board significantly reduced a sagittal plane knee displacement by approximately 15° during a drop-landing task. This study also observed a significant reduction of the sagittal plane knee displacement between NB [*M* (*SD*) = 65.7 (11.4)°] and HTB [*M* (*SD*) = 60.4 (9.7)°] conditions. However, it should be noted that the sagittal plane ankle displacement in this study did not attribute a reduction of peak dorsiflexion angle during the drop-landing task [NB: *M* (*SD*) = 26.7 (4.6)°, LTB: *M* (*SD*) = 28.5 (11.1)°, HTB: *M* (*SD*) = 26.4 (10.6)°]. Even though the static ankle ROM measurement showed significant decrease by EAS application (Table 2), we observed that the ankle displacement ascribed to the decreased plantar flexion angle at the initial foot contact [NB: *M* (*SD*) = 35.8 (12.3)°, LTB: *M* (*SD*) = 28.5 (11.1)°, HTB: *M* (*SD*) = 26.4 (10.6)°]. These results contradict previous studies[4,141] that reported the reduction of ankle ROM, specifically dorsiflexion, intensified the MKD during squatting exercises. We still agree with the fact that in those studies, the ankle displacement was mechanically controlled either to increase[4] or to decrease[141] the available ankle ROM. Therefore, the magnitude of MKD during a drop-landing task appears to depend on available static ankle ROM or dynamic ankle displacement. An additional correlation calculation showed that tMKD was inversely correlated with both knee extended ankle ROM ($r = -.45$, $P < .001$) and knee flexed ankle ROM (r = -.53, $P < .001$).

The importance of ankle displacement, specifically plantar flexion, was reported by Self and his colleagues.[3] When comparing two landing techniques during a vertical drop-landing from a 30cm height, stiff knee landing with exaggerated (greater) plantar flexion and stiff-knee landing with natural (less) plantar flexion, the ankle plantar flexion at initial foot contact was 33° and 45°, respectively. Their results showed the peak vGRF was approximately 18% smaller in the landing with greater plantar flexion.[3] These results were consistent with the findings of our study. The plantar flexion angle with NB at the initial foot contact in our study showed similar angle to Self et al.[3], and HTB condition demonstrated approximately 6% greater vGRF compared to NB during the drop-landing task. The difference in the magnitude of vGRF may attribute to the tasks performed and other joint displacement angles. Self and his research team[3] reported that participants performed vertical drop-landing tasks, and the knee displacement (M = 28.6°) was controlled in both landing conditions, while we did not control the knee angles during the drop-landing task [NB: M (SD) = 65.7 (11.4)°, LTB: M (SD) = 63.0 (10.0)°, HTB: M (SD) = 60.4 (9.7)°]. The smaller difference in vGRF compared to Self et al.[3] may be caused by the knee-flexion displacement. Yet, Hewett et al.[179] reported that ACL injured individuals had 20% greater GRF compared non injured individuals; that is, increased vGRF by 6% may not be enough to increase ACL strain. We found that sagittal plane knee displacement was significantly correlated with ankle displacement (r = .51, P < .001) and vGRF (r = -.65, P < .001). The greater the lower extremity sagittal plane joint displacement indicated the body had longer landing phase or greater angular velocity at a joint; as a result, the muscles of the lower extremities generate counter force for longer duration.[5] Greater knee flexion angle at the foot contact and subsequent increasing sagittal plane knee displacement were inversely correlated with peak GRF during landing tasks.[181] Therefore, the correlation between sagittal plane ankle displacement and sagittal plane knee displacement during a drop-landing task provided longer duration for the body to attenuate vGRF. By applying EAS, ankle ROM and resultant sagittal plane ankle displacement decreased. This limited ankle ROM was correlated with smaller knee displacement; hence, it is reasonable that greater vGRF was observed with EAS application.

Surprisingly, neither of tMKD or vGRF was correlated with knee-flexed isokinetic plantar flexor strength. In the second hypothesis, we believed that the importance of the gastrocnemius would become less in a landing phase; in contrast, the soleus would play more important role during a landing task because the knee joint would be flexed up to 90° during a landing task. Yet, the second hypothesis was not supported in this study. Although the importance of plantar flexors are proposed by Self et al.[3] and DeVita et al.[5] in landing tasks, controversial results were reported depending on a task performed. When comparing soft landing (greater knee displacement) and stiff landing (less knee displacement), studies found that the role of plantar flexors are noticeable,[3,5] while Yeow et al.[178] reported that the contribution of the plantar flexor muscles were trivial compared to the knee and the hip joint during a landing task. The discrepancy in the importance of plantar flexors seems to attribute the difference of landing techniques. When the knee displacements were controlled (stiff or soft landing depending on the knee flexion angle), the contribution of the plantar flexor muscles increased because the knee joint did not attenuate the landing force.[3,5,6] We did not control the sagittal plane knee flexion ROM or displacement; therefore, it is reasonable that the contribution of the plantar flexor might have been trivial to absorb the landing force in this study. Still, we found the vGRF was greater in EAS group compared to NB group possibly because the EAS application altered the sagittal plane ankle joint displacement and subsequent knee joint displacement. Our finding also did not show a correlation between isokinetic plantar flexors and tMKD. This no correlation indicated that the contribution of the plantar flexor was minor in a drop-landing task probably because the sagittal plane knee displacement was not controlled and compensated the limited ankle displacement in this study. Moreover, the muscles that mainly counteracted the knee valgus movement were the knee musculature located on the medial half of the knee joint, such as gracilis, medial hamstrings, vastus medialis, rectus femoris, and medial gastrocnemius. However the plantar flexors, especially gastrocnemius, did not contribute to resist against valgus moment.[182] Therefore, the vGRF needs to be absorbed by the knee and ankle joints in conjunction with the hip joint that is depending on the landing techniques.[115,178,183]

Limitations

This study had several limitations. We recruited only female subjects who were not participating in intercollegiate sports but were physically active college-age young adults. Non-contact ACL injury occurs more often in athletic settings, so the results of this study can only be generalized to physically active college-age females. Future studies also need to recruit male subjects and different age group subject to confirm this study. The second limitation is that because we did not use three-dimension motion analysis, the MKD measured in two-dimension kinematic assessment is assumed to represent knee valgus.[45] However, we can not specify the angle of knee valgus in each knee because the MKD simply represent the proportion of the distance between the centers of the patellae relative to the distance between the anterior superior iliac spines. In addition to the assumption of the MKD, we were also not able to determine the joint movements in the transverse plane due to two-dimension kinematic analysis. These analyses need to be done with more sophisticated equipment.

Conclusions and Clinical Implications

It is believed that greater knee valgus angle and greater vGRF increased the risk of non-contact ACL injury.[37] The sagittal plane joint motions altered the frontal plane knee motion and landing force regardless of the isokinetic soleus strength. HTB intensified tMKD and vGRF due to reduced ankle displacements during a drop-landing task even though the effect sizes were medium and small, respectively. The ankle ROM was limited by prophylactic lace-up ankle braces, and its effect size was also large. While only 11% preferred the HTB condition over LTB, most of subjects commented that the LTB felt “snug.” Therefore, health care professionals who apply ankle taping or bracing on physically active individuals need to be aware of not applying EAS with excessively restricting the ankle sagittal plane ROM. The primary finding of this study was that the prophylactic lace-up ankle braces significantly increased tMKD along with vGRF during a landing task, but the magnitude of ACL strain was not measured. It is recommended that those who regularly wear EAS during activities need to train with EAS when they practice landing techniques. Hence, future studies should aim at measuring frontal plane knee displacement and GRF during a landing task should be considered to investigate the magnitude of tMKD and vGRF to sufficiently increase the stress on the ACL to tear.

Table 1. Orders of Ankle Bracing Conditions and Landing Tasks for Each Participant.

Subject #	Ankle condition 1	Ankle condition 2	Ankle condition 3	Landing task 1	Landing task 2
1	NB	LTB	HTB	DJL	FJL
2	LTB	HTB	NB	DJL	FJL
3	HTB	NB	LTB	DJL	FJL
4	NB	LTB	HTB	FJL	DJL
5	LTB	HTB	NB	FJL	DJL
6	HTB	NB	LTB	FJL	DJL
7	NB	LTB	HTB	DJL	FJL
8	LTB	HTB	NB	DJL	FJL
9	HTB	NB	LTB	DJL	FJL
10	NB	LTB	HTB	FJL	DJL
11	LTB	HTB	NB	FJL	DJL
12	HTB	NB	LTB	FJL	DJL
13	NB	LTB	HTB	DJL	FJL
14	LTB	HTB	NB	DJL	FJL
15	HTB	NB	LTB	DJL	FJL
16	NB	LTB	HTB	FJL	DJL
17	LTB	HTB	NB	FJL	DJL
18	HTB	NB	LTB	FJL	DJL
19	NB	LTB	HTB	DJL	FJL

Table 2. Means (*SD*) of Ankle Displacement, Knee Displacement, Total Medial Knee Displacement, and Vertical Ground Reaction Force

	NB	LTB	HTB	*P* value (ANOVA)
Ankle Displacement (°)	62.5 (13.2)	51.6 (10.7)	48.2 (10.8)	< 0.001
tMKD (%)	20.1 (15.9)	26.2 (21.2)	35.1 (24.6)	< 0.001
vGRF (N/BW)	2.61 (0.43)	2.68 (0.40)	2.76 (0.40)	0.007

Abbreviation: NB = No bracing, LTB = Low-tension bracing, HTB = High-tension bracing

Table 3. Means (*SD*) of Ankle Range of Motions for Ankle Brace Conditions and Measurement Procedures

	NB			LTB			HTB			*P* value		
	PF	DF	Total ROM	PF	DF	Total ROM	PF	DF	Total ROM	PF	DF	Total ROM
Knee Extended (°)	44.12 (12.2)	10.42 (5.3)	54.54 (12.1)	33.4 (9.9)	10.1 (4.7)	43.5 (10.9)	22.0 (6.5)	6.8 (6.0)	28.8 (9.9)	<.001	<.001	<.001
Knee Flexed (°)	43.7 (13.6)	16.1 (5.3)	59.7 (14.3)	31.1 (11.0)	14.1 (4.7)	45.2 (11.4)	22.5 (7.3)	11.6 (5.5)	34.1 (9.4)	<.001	<.001	<.001
Lunge DF (°)		43.9 (6.5)			40.7 (4.2)			38.0 (4.3)			<.001	
P value	0.17	<.001	<.001	.02	<.001	.05	.53	<.001	<.001			

Abbreviations: NB = No bracing, LTB = Low-tension bracing, HTB = High-tension bracing, PF = Plantar flexion, DF = Dorsiflexion, Total ROM = PF + DF

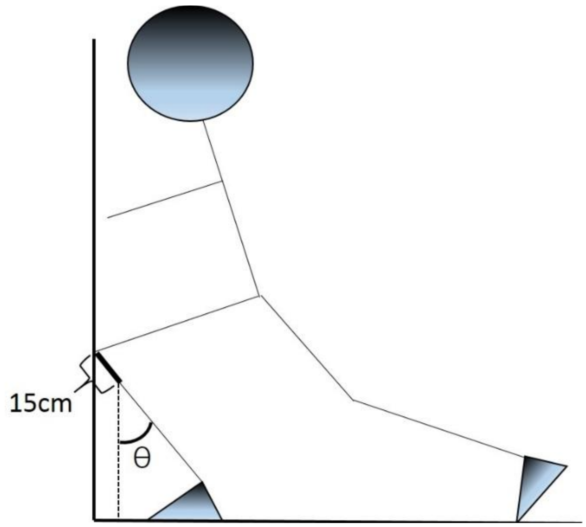

Figure 1. Procedure of Weight-bearing Ankle Dorsiflexion Measurement.
1) Participant lunged forward until the ipsilateral knee was in contact with the wall. 2) The distance between the foot and the wall was adjusted to find the point where the knee could touch the wall without lifting the heel off the floor. 3) When participants reached the maximum dorsiflexion angle, the investigator recorded the tibial angle by a fluid filled inclinometer (Baseline Bubble; Fabrication Enterprises, Inc., White Plains, NY).

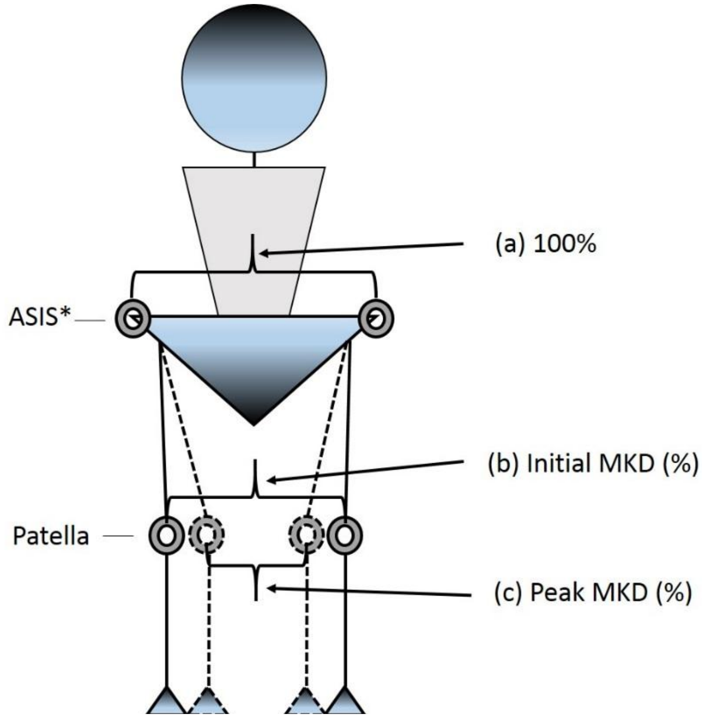

Figure 2. Definition of Total Medial Knee Displacement (tMKD).
The total medial knee displacement (tMKD) is calculated by (b) – (c).
* ASIS = Anterior superior iliac spine. (a) Normalized distance between ASISs (= 100%). (b) Distance between the patellae relative to (a) at initial foot contact. (c) Distance between the patellae relative to (a) at maximum knee flexion.

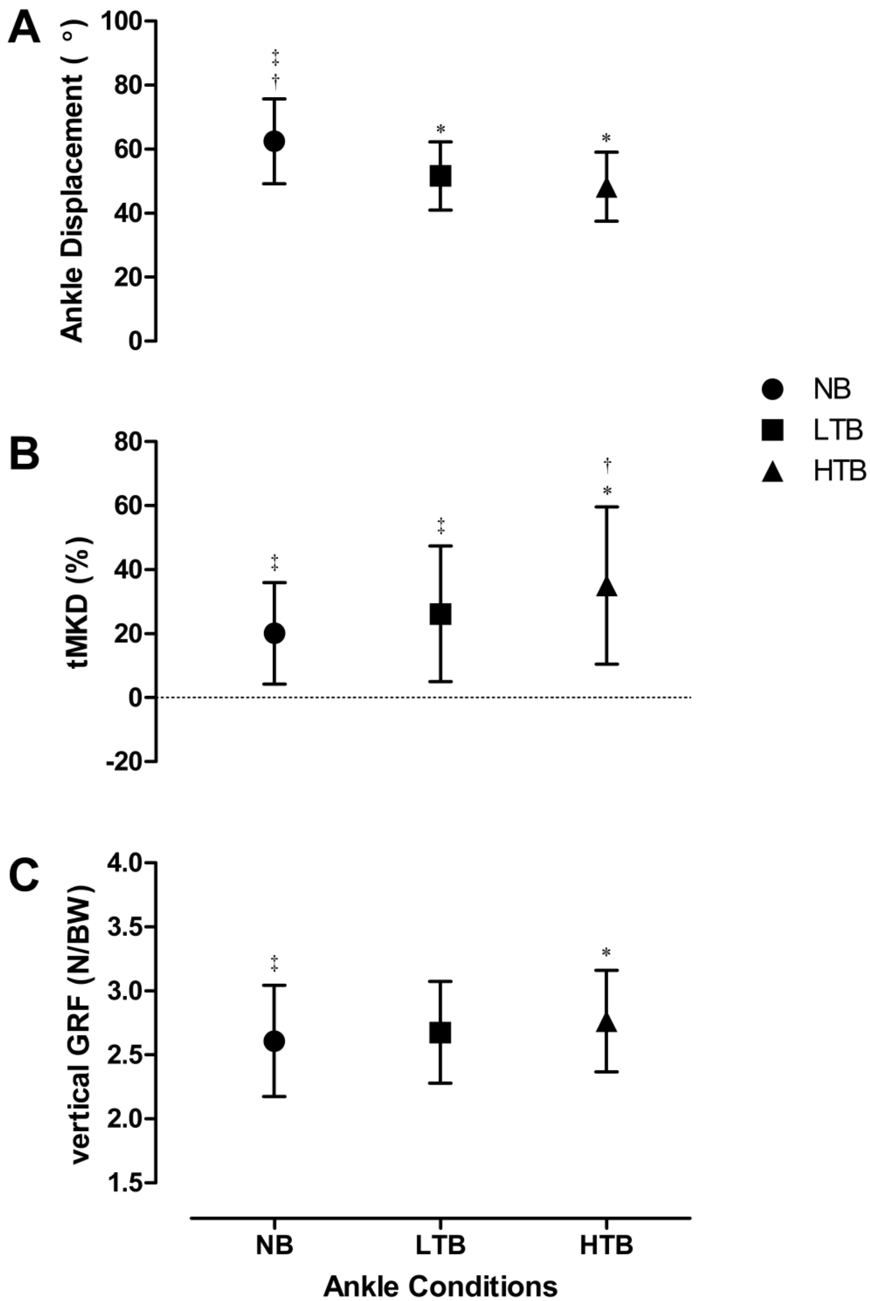

Figure 3. External Ankle Supports Altered Landing Kinematics and Kinetics.
Ankle displacement (A), total medial knee displacement (tMKD, B), and vertical ground reaction force (vGRF, C) in three ankle brace conditions (no-bracing = NB, low-tension bracing = LTB, high-tension bracing = HTB). * = significantly differs from NB. † = significantly differs from LTB. ‡ = significantly differs from HTB.

PAPER TWO: ANKLE BRACING ALTER LANDING MECHANICS DURING DROP-LANDING AND FORWARD-LANDING TASKS

Abstract

Context: Landing with greater knee valgus angle and ground reaction force (GRF) increases anterior cruciate ligament (ACL) strain in a deceleration phase. Sagittal plane ankle movement is important not only to absorb GRF, but also to reduce medial knee displacement (MKD) during landing tasks. Although external ankle supports (EAS) are commonly used to limit frontal plane ankle range of motion (ROM) in athletic settings, EAS may also restrict sagittal plane ankle ROM. Drop-jump landing (DJL) and forward-jump landing (FJL) are commonly used to evaluate the risk of non-contact ACL injury; however, differences in landing kinematics and kinetics of the two tasks are not well studied. **Objective:** To evaluate the effect of EAS on kinematics and kinetics in DJL and FJL tasks. **Design:** Crossover study. **Setting:** Controlled laboratory environment. **Patients or Other Participants:** Nineteen physically active females [*M* (*SD*): age = 20.2 (1.1) years, height = 170.0 (7.15) cm, mass = 65.7 (8.0) kg]. **Intervention(s):** Participants performed a DJL and FJL tasks under three ankle bracing conditions: no bracing, low-tensioned bracing, and high-tensioned bracing. **Main outcome Measure(s):** Ankle displacement, knee displacement, peak MKD, vertical GRF (vGRF), and posterior GRF (postGRF). **Results:** Significant effects of EAS were found in ankle displacement ($F_{2,36}$ = 46.73, P < .001), knee displacement ($F_{36,2}$ = 23.46, P < .001), and peak MKD ($F_{2,36}$ = 56.45, P < .001). Landing tasks affected only postGRF ($F_{1.18}$ = 440.65, P < .001). Significant interaction between ankle brace conditions and landing tasks ($F_{36,2}$ = 6.84, P < .01) was observed in vGRF. **Conclusions:** Restricting sagittal plane ankle ROM increased peak MKD and postGRF in both landing tasks, and may raise the risk of non-contact ACL injury. Therefore,

healthcare professional should be careful not to excessively limit the ankle motion when applying EAS.

Introduction

Knee ligamentous injuries including anterior cruciate ligament (ACL) rupture were more common among female athletes.[10] Two major tasks performed during non-contact ACL tear were cutting and landing tasks at a deceleration phase.[28] Landing with increased knee valgus angle and greater ground reaction force (GRF) produced greater ACL strain during a deceleration task.[2] Therefore, greater knee valgus angle could lead to non-contact ACL injury due to subsequent loading of the ACL during a landing or deceleration task.

The foot is the first part of the body to make contact with a landing surface, and the foot position at the initial contact could decide the magnitude of sagittal plane ankle joint displacement.[5,6] A video analysis study[35] reported that ACL injured subjects showed less plantar flexion of the ankle at initial contact phase of landing. Hence, the ankle joint plays an important role in decelerating and changing a direction to absorb impact from the ground or a playing surface. Hewett's[35] study also revealed the commonly observed mechanism of non-contact ACL injury was "valgus collapse" of the knee joint. The valgus collapse is a combination of knee abduction, external rotation of the tibia, and internal rotation of the femur.[28,35,37,80] In this study, medial knee displacement (MKD) represents knee valgus collapse, and it is operationally defined as apparent movement of the knee joint in the frontal plane and involved the hip, knee, and ankle joints' sagittal, frontal, and transverse plane movements.[47] Although the two-dimension analysis could not identify rotational movement, a previous study[45] reported two-dimension analysis can be used to distinguish excessive knee valgus, a movement in the frontal plane.

Sagittal plane ankle and knee joint kinematics appeared to affect frontal plane knee kinematics.[39,40] Less ankle dorsiflexion and smaller plantar flexor strength increased the MKD during a squat exercise that simulated descending phase of a landing task.[4] Hence, MKD could be increased by limited ankle ROM, particularly dorsiflexion during a landing task. Ankle displacement was important for plantar flexors to generate a counter force during a landing task. The landing impact forced the ankle joint into dorsiflexion, and the eccentrically contracting plantar flexor muscles counteracted the dorsiflexion to absorb GRF during a toe landing task.[108] When a landing with greater plantar flexion was compared to a landing without plantar flexion, a greater ankle plantar flexion prior to the foot contact exhibited smaller vertical ground reaction force compared to less ankle plantar flexion.[7] Therefore, greater plantar flexion and subsequent greater ankle displacement could be important to maximally absorb vertical landing force.

Tasks of landing could influence the kinematics and kinetics. Two commonly used landing tasks to assess non-contact ACL injury risk were drop-jump landing (DJL),[3,5,86,119,124] and forward-jump landing (FJL).[113,124] A study comparing the two landing tasks concluded that the two tasks were different. In test settings, forward-jump landing involved concentric (in taking-off phase) and eccentric (in landing phase) muscle contraction during the task; on the other hand, drop-landing involved only eccentric (in landing phase) muscle contraction.[82] Although two different types of landing tasks have been examined to assess the risk of non-contact ACL injury, a comparison of the landing types, DJL and FJL, in terms of the kinematics and kinetics of ankle and knee joints have not been studied. Therefore, this study compared the two types of landing tasks in terms of ankle and knee displacement during the two types of landing tasks.

In addition, external ankle supports (EAS), including ankle taping and bracing, have been widely used in various physical activities. Although the main purpose of EAS is to limit the frontal plane ankle motion and to reduce the risk of ankle sprain, the application of EAS also restricts the sagittal plane ankle motions.[8,41] This restricted sagittal plane ankle could contribute to alter the landing kinematics and kinetics.[5] Restriction of the ankle dorsiflexion reduced peak knee flexion angle and increased MKD during a squatting task.[141] Limiting dorsiflexion in a landing task, then, could increase the risk of non-contact ACL injury due to increasing knee valgus angle. Limited dorsiflexion during a deceleration phase of activities appeared to increase MKD,[4,141] and this greater MKD could increase the risk of the ACL injury. Therefore, the purpose of this study was to assess whether the limited ankle ROM by application of EAS and landing tasks would similarly change landing kinematics and kinetics during DJL and FJL tasks. Two hypotheses were investigated. First, there would be difference in sagittal plane ankle displacement, sagittal plane knee displacement, peak MKD, vertical GRF (vGRF), and posterior GRF (postGRF) during landing tasks among ankle brace conditions controlled by EAS application. Second, there would be difference in sagittal plane ankle displacement, sagittal plane knee displacement, peak MKD, vGRF, and postGRF between DJL and FJL tasks.

Methods

Participants

Nineteen physically active and healthy college aged female subjects [*M* (*SD*): height = 170.0 (7.15) cm; mass = 65.7 (8.0) kg, age = 20.2 (1.1) years, 11.4 (9.2) days post menstrual period] without history of lower extremity injury completed the test protocol. The inclusion criteria were participation in physical activity at least twice per week and minimum of 30 minutes per session, no current neuromuscular dysfunction, no signs or symptom of inflammation, no history of surgeries within six months in the lower extremities. The participants were recruited from a local university and completed a demographic form, and signed an informed consent prior to being tested. The study was approved by the universities Institutional Review Board.

Ankle Range of Motion Measurement

After obtaining the informed consent, demographic and anthropometric information, a 5-minute warm-up exercise was performed. Ankle ROM was measured in non-weight bearing and weight bearing positions. The non-weight bearing active ROM of the ankles was measured using a plastic goniometer (Patterson Medical/ Sammons Preston; Bolingbrook, IL) in two positions. First, the subject was in long-leg sitting position; second, the subject was sitting with 90° hip and knee flexion. Two measurements were taken by the same researcher and averaged for analyses.

In weight bearing position, lunge dorsiflexion was measured by determining the angle of the lower leg relative to the vertical line.[158] Because the torque applied to the ankle joint at the weight bearing position was greater than that applied by an examiner's hand, the advantage of this method was that the measurement could more represent the ankle dorsiflexion during functional tasks, such as running, jumping, and landing.[158] First, a single strip of tape was aligned on a wall perpendicularly and on the floor 90° to the wall. Second, two landmarks on the skin were marked by a felt pen. One landmark was at 15 cm inferior to the tibial tuberosity on the anterior boarder of the tibia. The other was a line perpendicularly bisecting the posterior calcaneus. Third, participants positioned the foot on the tape with the big toe and the line on the calcaneus. Fourth, participants lunged forward until the ipsilateral knee was in contact with a vertical tape on the wall. The distance between the foot and the wall was adjusted to find the point where the knee could touch the wall without lifting the heel off the floor. The heel was held by the investigator in order to maintain heel contact. Lastly, when participants reached the maximum dorsiflexion angle, the investigator placed the fluid filled inclinometer (Baseline Bubble; Fabrication Enterprises, Inc., White Plains, NY) on the mark placed on the tibia and recorded the tibial angle. During the lunge dorsiflexion, pronation or supination of the foot was allowed. Two measurements were taken by the same researcher and averaged for analyses. Although this method did not measure a specific joint ROM but measured a combined movement at talocrural, subtalar, and inter-tarsal joint, both intra-tester reliability and inter-tester reliability were excellent.[158]

External Ankle Supports Application

Prophylactic lace-up ankle braces (ASO® Ankle Stabilizer; Medical Specialties, Inc., Charlotte, NC) were fit on both the subject's both ankles as instructed by the manufacturer. During the brace application, participants were sitting in a chair and the lower leg was kept in perpendicular to the floor so that the ankle joint was maintained in 90°. Each lace was tied with two lace tensions at 2.5 – 3.0 kg (lower-tension brace, LTB) and at 7 – 7.5 kg (higher-tension brace, HTB). The lace-up tensions were measured by a hand-held digital scale (Rapala VMC, Vaaksy, Finland). Each lace was held between the target tensions for one second at 90° angle from the surface contour of the ankle braces.

Jump-Landing Tasks

Participants were instructed to wear athletic shirt, shorts, and low cut sports shoes. For frontal plane kinematic recordings, markers were placed at both left and right side of the body; at 2 cm anteriorly to the acromioclavicular joint, at the anterior superior iliac spine, at the center of patella, and at the middle of the distal lower extremity (5cm superior to the ankle joint). For sagittal plane kinematic recordings, markers were placed only on the right side of the body; at 2 cm lateral to the acromioclavicular joint, at the greater trochanter, at the center of the knee joint line, and at the lateral malleolus. Two digital cameras (EX-FH20; Casio, Inc., Tokyo, Japan) were used to record landing kinematics. Digital cameras were mounted on a tripod 73 cm from the floor. One tripod was placed 175 cm in front of the center of the force plate where a participants aimed to land. This digital camera recorded peak MKD in frontal plane. Another tripod with a digital camera mounted on was positioned at 175 cm on the right side of the force plate. This second digital camera recorded ankle and knee displacement in sagittal plane. The force platform (AccuPower; AMTI, Watertown, MA) was embedded to the testing surface. The position of the 33-cm box was adjusted depending on the task performed.

Prior to a trial jump, participants were allowed to practice each jump-landing task until they felt comfortable to perform the landing tasks. Although the investigator verbally explained the sequence of the landing tasks, no specific instruction was provided regarding the landing tasks to minimize the coaching effect on the participant's natural landing strategies. Minimum of one complete jump-landing sequence was verified by the investigator before testing trials to make sure understanding of the each task. The drop-jump landing (DJL) protocol followed the previously used protocol[47] with some modifications. A 33cm box was placed 5cm from the back edge of the platform, which was 35 cm from the landing target on the platform. The five-step DJL sequence was as follows: (1) participant extended the right leg in forward with approximately 30° of hip flexion; (2) participant took off with the left foot without jumping up movement; (3) participant landed on the force platform immediately below the extended right foot; hence the task was relatively vertical landing; (4) participant performed a maximum vertical jump immediately after two-foot landing; and (5) participant landed back on the platform.

The forward-jump landing (FJL) protocol also followed the previously used protocol[46] with a modification. The five-step FJL sequence was as follows: (1) participant jumped off with two feet from the 33cm box located at 50% of participant's height to the center of the force plate; (2) participants jumped forward, but not vertically, to reach the center of the force platform; (3) participant landed on the center of the platform with both feet; (4) participants performed a maximum vertical jump immediately after two-foot landing; and (5) participants landed back on the platform.

Data Collection

Three ankle brace conditions (no prophylactic ankle bracing = NB, low-tension ankle bracing = LTB, and high-tension ankle bracing = HTB) and two landing tasks (drop-jump landing = DJL and forward-jump landing = FJL) were counterbalanced and alternated, respectively for all participants. The digital cameras captured the landing tasks at a sampling rate of 210 Hz. The data stored in memory cards from each camera were transferred to a laptop computer for kinematic analyses using Dartfish Motion Analysis Software (version 10.0; Dargfish®, Fribourg, Switzerland). Force plate data were collected at a sampling rate of 400 Hz. The force platform was connected to a laptop computer so that the GRF data were stored in the computer for kinetic analyses by using computer software (AccuPower version 1.5; Athletic Republic, Fargo, ND).

Data Analysis

Kinematic measurements

Peak MKD was operationally defined as the ratio of distance between the center of left and right patellae relative to the distance between the left and right anterior superior iliac spines. The distance of those landmarks was generally dependent on participant's body size; therefore, the ratio was considered as better measurement than the actual distance between the patellae. Therefore, in this study, smaller peak MKD indicates greater knee valgus that represents greater knee valgus angle. This analysis was performed on a computer screen with the Dartfish Video Motion Analysis software. Values from the two measurements in each condition were averaged for statistical analyses.

The ankle displacement was defined as the total angle between the foot contact and the maximum dorsiflexion during a landing task. Sagittal plane ankle angles were measured only on the right side of the body at two instants during an initial landing from the take off box. The first instant was at the foot contact when any part of the right foot made contact with the surface of the force platform. The second instant was measured at maximum dorsiflexion of the ankle. Total angle between the two instants were calculated as a displacement. Ideally, the anatomical references for ankle angle measurement were the shaft of the lower extremity and the shaft of the 5th metatarsal. The first line was the shaft of the lower extremity that is referenced between the center of the knee joint line and the lateral malleolus. Because the 5th metatarsal was covered by the shoe, the landmark was assumed to be parallel to the sole of the shoe. Therefore, the second line was drawn from the lateral malleolus toward the toes that is parallel to the sole of the shoe. The knee angles were measured as the angle created by the two lines; between the lateral malleolus and the knee joint line and between the knee joint line and the greater trochanter of the femur. Values from the two trials in each condition were averaged for statistical analyses.

Kinetic measurements

Peak vGRF and peak posterior ground reaction force (postGRF) were obtained during a landing task. AccuPower software was used to analyze the GRFs. Raw GRF values were re-calculated relative to each participant's body weight (BW in N). Values from the two trials in each condition were averaged for statistical analyses. The GRF values were expressed as a two-feet landing force.

Statistical Analysis

The sample size was determined based on the pilot study in which the minimum expected difference and standard deviation (*SD*) in ankle dorsiflexion during a DJL task between NB condition and HTB condition were 3.47° and 3.9°, respectively. The means and *SD* for each dependent variable were calculated across the two trials for each ankle condition and landing task. Prior to evaluating the hypotheses, $ICC_{3,2}$ was calculated to check the reliability of kinematic dependent variables. Independent variables were ankle brace conditions and landing tasks. Dependent variables were ankle displacement, knee displacement, peak MKD, vGRF, and postGRF. The effect of ankle brace conditions and landing tasks on ankle displacement, knee displacement, peak MKD, vGRF, and postGRF were evaluated by performing a repeated measure 3 (ankle brace conditions) x 2 (landing tasks) MANOVA because these dependent variables were believed to be associated. Following the assumption checking procedures, interaction was evaluated. When interaction was not significant, separate repeated measure ANOVAs with Tukey post hoc test at each level of factors were conducted to examine the main effects. Following conducting necessary ANOVAs, η^2 of effect size was calculated to assist clinical interpretation. The effect size η^2 was defined as follows; trivial < 0.01, small < 0.06, medium < 0.14, and 0.14 ≤ large.[177] The α level was set a priori at ≤ .05 for all statistical analyses. Statistical Package for the Social Sciences (version 20.0, SPSS Inc, Chicago, IL) was used to analyze the data.

Results

Two hypotheses were investigated. First, there would be difference in sagittal plane ankle displacement, sagittal plane knee displacement, peak MKD, vGRF, and postGRF during landing tasks among ankle brace conditions controlled by EAS application. Second, there would be difference in sagittal plane ankle displacement, sagittal plane knee displacement, peak MKD, vGRF, and postGRF between DJL and FJL tasks. Before MANOVA was conducted, $ICC_{3,2}$ in each dependent variable was calculated as reliability testing. Intraclass correlation coefficient for each dependent variable are as followings: ankle displacement (DJL: $ICC_{3,2} = 0.98$, FJL: $ICC_{3,2} = 0.94$), peak MKD (DJL: $ICC_{3,2} = 0.96$, FJL: ICC3,2 = 0.97), vGRF (DJL: $ICC_{3,2} = 0.83$, FJL: $ICC_{3,2} = 0.77$), and postGRF (DJL: $ICC_{3,2} = 0.71$, FJL: $ICC_{3,2} = 0.60$). MANOVA was conducted to assess if there were significant effects of ankle brace conditions (NB, LTB, and HTB) and landing tasks (DJL and FJL) on ankle displacement, knee displacement, peak MKD, vGRF, and postGRF. The assumptions of MANOVA were checked and met. A significant effect was found in ankle brace conditions (Wilks' $\Lambda = 0.740$, $F_{12,206} = 3.38$, $P < .001$) and landing tasks (Wilks' $\Lambda = 0.639$, $F_{6,103} = 11.75$, $P < .001$), but not in the interaction (Wilks' $\Lambda = 0.947$, $F_{12,206} = 0.57$, $P = .83$). Because omnibus F value of MANOVA was significant in the factors, we conducted separate 3 (ankle brace conditions) x 2 (landing tasks) repeated measures ANOVA on each dependent variable.

Follow up 2 x 3 repeated measures ANOVAs with Tukey post hoc test was conducted to assess whether there were difference in ankle displacement, knee displacement, peak MKD, vGRF, and postGRF. Results for ankle displacement indicated that no interaction was found between ankle brace conditions and landing tasks ($F_{2,36} = 0.63$, $P = .54$). Significant effects of ankle brace conditions were found ($F_{2,36} = 46.73$, $P < .001$, observed power > 0.99, $\eta^2 = 0.16$), but no effect was observed between landing tasks ($F_{1,18} = 0.36$, $P < .561$). Post hoc test results indicated that ankle displacements with both LTB and HTB were significantly less than NB ($P < .05$), but no difference was observed between LTB and HTB (Figure 4). Hence, the first hypothesis was supported. Means and *SD* of ankle displacement are shown in Table 4.

Result for knee displacement revealed that no interaction was observed between ankle brace conditions and landing tasks ($F_{2,36} = 0.72$, $P = .49$). EAS application significantly changed the knee displacement during landing tasks ($F_{36,2} = 23.46$, $P < .001$, observed power > 0.99, $\eta^2 = 0.05$). Post hoc test indicated that the knee displacement was significantly smaller in LTB compared to NB condition ($P < .05$), and it was significantly smaller in HTB compared to LTB condition ($P < .05$) in both landing tasks. Also, the knee displacement was also greater in DJL compared to FJL ($F_{18,1} = 32.71$, $P < .001$, observed power > 0.99, $\eta^2 = 0.06$) in all ankle brace conditions (Figure 4). Hence, the first hypothesis was supported. Means and *SD* of knee displacement are shown in Table 4.

Results for peak MKD revealed that the change was significant in ankle brace conditions ($F_{2,36} = 56.45$, $P < .001$, observed power > 0.99, $\eta^2 = 0.05$) but not in landing tasks ($F_{1,18} = 1.76$, $P = .20$). Also, no interaction was observed between ankle brace conditions and landing tasks on peak MKD ($F_{2,36} = 0.11$, $P = .89$). Post-hoc test results indicated that peak MKD with LTB was significantly smaller than NB ($P < .05$) but greater than HTB, and peak MKD with HTB was significantly smaller than NB and LTB ($P < .05$) (Figure 4). Hence, the first hypothesis was supported. Means and *SD* of peak MKD are shown in Table 4.

A two-way repeated measures ANOVA for vGRF showed a significant interaction between ankle brace conditions and landing tasks ($F_{36,2} = 6.84$, $P < .01$), and this indicated ankle brace conditions changed vGRF differently in DJL and FJL (Figure 5). Means and *SD* of vGRF are shown in Table 4. A follow up repeated measures ANOVA for ankle brace conditions reported that a significant effect of ankle brace conditions on vGRF was observed ($F_{36,2} = 5.72$, $P < .007$, observed power > 0.84, $\eta^2 = 0.03$), and post hoc Tukey multiple comparison test showed vGRF was significantly greater in HTB condition compared to NB condition ($P < .05$) during DJL task. Another follow up repeated measures ANOVA for ankle brace conditions in FJL was also performed to examine the effect of ankle brace conditions on vGRF. Result showed the effect of ankle brace conditions was significant ($F_{36,2} = 4.34$, $P = .02$, observed power = 0.54, $\eta^2 = 0.03$) in FJL. Post hoc Tukey multiple comparison test reported that during FJL, vGRF was significantly greater in NB than HTB condition ($P < .05$). Therefore, both the first and second hypotheses were supported in regard to vGRF.

Results for postGRF indicated that a significant change in postGRF was observed between landing tasks ($F_{1.18} = 440.65$, $P < .001$, observed power > 0.99, $\eta^2 = 0.75$), but not among ankle brace conditions ($F_{2,36} = 3.00$, $P = .06$). No interaction was observed between ankle brace conditions and landing tasks on postGRF ($F_{2,36} = 0.60$, $P = .55$). Post hoc Tukey multiple comparison test revealed that postGRF in FJL was significantly greater than postGRF in DJL ($P < .05$) (Figure 4). Thus, only the second hypothesis was supported regarding postGRF. Means and *SD* of postGRF are shown in Table 4.

Discussion

Although EAS is used to limit the frontal plane ankle ROM to reduce the risk of ankle sprain, the application of EAS also restricts the sagittal plane ankle ROM.[8,41] This restricted sagittal plane ankle ROM appeared to alter the landing kinematics and kinetics.[5] Smaller dorsiflexion during a deceleration phase of activities was found to increase subsequent MKD,[4,141] and this limited dorsiflexion in a landing task could increase the risk of non-contact ACL injury due to increasing MKD or knee valgus. Although, two different types of landing tasks, DJL and FJL, have been used to assess the risk of non-contact ACL injury, the kinematics of ankle and peak MKD have not been studied in two different types of landing tasks to date. The purpose of this study was to assess whether the limited ankle ROM by application of EAS and the landing tasks would change landing kinematics and kinetics.

In this study, the laces of the ASO® prophylactic ankle braces were controlled in two tensions. The laces were tied with a tension between 3 and 5 kg in LTB condition. A majority (17 out of 19) of subject commented that LTB condition was felt "snug." On the other hand, in HTB condition, the laces were tied with a tension between 7 and 8 kg. Only small portion (2 out of 19) of subject preferred the HTB conditions. The effect of ASO prophylactic ankle brace to restrict the sagittal plane ankle displacement during DJL was comparable to the results of a previous study.[134] DiStefano and colleagues[134] reported that the means (*SD*) of ankle joint displacement during a jump-landing task were 59.3 (14.3)° and 56.5 (12.5)° without and with EAS, respectively.

We found that restricted sagittal plane ROM at the ankle joint reduced the ankle displacement, decreased knee displacement, increased peak MKD in both DJL and FJL similarly. The restricted sagittal plane motion at the ankle joint affected vGRF differently depending on the task performed. The postGRF was also greater in FJL compared to DJL; however, the ankle brace conditions did not change postGRF in either DJL or FJL.

Among those findings, the primary finding was that restricted sagittal plane ankle displacement by EAS application increased peak MKD in both DJL and FJL; in other words, limited ankle ROM by EAS increased MKD regardless of landing tasks. This finding was consistent with previous studies[4,141] measured frontal plane knee motion during a two-leg squat.

Macrum et al.[141] found that restricted ankle dorsiflexion by placing a wedge under the forefoot lead to greater MKD excursion compared to non-wedge condition. Bell and colleagues[4] observed that subjects with greater MKD showed less ankle dorsiflexion measured in knee flexed position compared to control subjects. These findings indicated restricting lower extremity sagittal plane joint displacement could increase the frontal plane knee displacement. During a deceleration phase of landing, the center of mass must be lowered by flexing the lower extremity joints in the sagittal plane. Limiting the sufficient ankle dorsiflexion could induce pronation at the subtalar joint and subsequent internal rotation of the tibia and knee valgus.[39,180] We also found that EAS application significantly decreased the knee flexion displacement in conjunction with an increased peak MKD during landings compared to NB condition. Thus, not only sagittal plane ankle displacement, less flexion in the knee were associated with greater peak knee valgus angle.[40,113] Because the MKD or knee valgus during a dynamic task was proposed to be one of the mechanisms of ACL injury,[35,99] it is plausible EAS application could increase the risk of non-contact ACL injury.

Although EAS application significantly decreased the sagittal plane ankle ROMs measured in knee extended and knee flexed positions (Table 5), no significant correlation was observed between ankle displacement and non-weight bearing ankle ROMs. This findings contradicted a previous correlational study in which extended knee ankle dorsiflexion ROM was significantly related to knee flexion displacement, but not related to MKD.[113] This contradiction could attribute to the different ankle ROM measurements. In our study, active ankle ROM was measured, while Fong et al.[113] used passive ankle ROM. Among measured ankle ROMs, the only correlation was found between ankle displacement and ankle dorsiflexion measured in the weight-bearing lunge position. During a landing, ankle joints are forced into dorsiflexion; thus, active ankle ROM might not have shown a significant correlation in this study.

In addition, our results showed a significant correlation between the sagittal plane ankle displacement and vGRF, which indicated the ankle joint could not have absorbed the vGRF effectively. According to the previous studies, [17,31] the less weight bearing dorsiflexion ROM produced greater stress on the proximal joint, and less ankle displacement was associated with greater vGRF during a drop-landing task. Moreover, extended knee passive dorsiflexion was also correlated with vGRF.[113] This study revealed that EAS decreased angular displacement at the ankle joint similarly in DJL and FJL but changed vGRF differently between two landing tasks. Regarding DJL, vGRF was significantly greater in HTB condition compared to NB condition. Therefore, EAS application not only increased peak MKD but also vGRF during drop-landing tasks due to limited ankle joint function to absorb impact from a landing surface.[35] In a correlational study,[34] ACL injured individuals demonstrated a correlation between vGRF and peak sagittal plane knee angle. Hence, our results indicated that excessively restricted sagittal plane ankle ROM may increase the risk of non contact ACL injury during a deceleration phase of landing tasks due to intensifying vGRF and peak MKD. Another related finding in our study was vGRF was significantly correlated with the following; ankle displacement (r = -0.35, P = .008), knee displacement (r = -0.65, P < .001), and hip displacement (r = -0.72, P < .001) during DJL task. This association was explained in other studies[5,6] that GRF must be absorbed by displacement of the ankle, knee, and hip joints and associated muscles. We also found a correlation between the ankle displacement and knee displacement (r = 0.51, P < .001) in DJL task. The landing load was decreased by the joints in the lower extremities, but joint displacement in a joint was affected by the displacements of other joints.[6,34,113] When one joint displacement is restricted, the other joints are expected to compensate the limited force absorbing ability. If the other joint does not perform the compensatory force reduction, the landing force would increase. Our result infers that displacement of both ankle and knee joints play important role in landing energy absorption in landing tasks, and the limited ankle ROM by EAS changed the landing style from natural landing to stiff landing. The stiff landing was characterized by more erect posture with less sagittal plane displacement during a landing.[5,34] The ankle joint absorbed approximately 80% of landing force when the knee and hip ROM was limited.[6] Hence, it is speculated that EAS application prevented the function of ankle and knee joints from absorbing landing energy and increased vGRF during DJL unless the hip displacement changed. This excessive vGRF was believed to be one of the mechanisms of the non contact ACL injury.[35] Another significant correlational result was observed between vGRF and peak MKD (r = -0.35, P = .007). The frontal plane energy dissipation was significantly greater in the hip and knee joint compared to the ankle joint regardless of the height of landing.[181] In our study, the vGRF was correlated with peak MKD (r = -0.35, P = .008). This speculated that the frontal plane displacements in the knee and hip joint also contribute to attenuate the landing impact during a drop-landing task.

The effect of EAS influenced vGRF differently in DJL and FJL, and a significant interaction was observed in this study. Greater vGRF was observed with EAS condition in DJL; in contrast, greater vGRF was observed with NB condition in FJL. The ankle displacement was significantly greater in NB compared to LTB and HTB conditions ($F_{36,2}$ = 18.12, P < .001, observed power = 0.99, η^2 = 0.12). Fong et al.[113] reported that extended knee dorsiflexion ROM was positively associated with knee displacement and negatively related to vGRF during a forward-landing task in which the landing target was set at 40% of the subject's height. Our result exhibited contradictory result regarding vGRF in which EAS not only significantly reduced the ankle displacement but also decreased the vGRF in FJL. Correlational statistics showed that vGRF in FJL was significantly associated with the knee displacement (r = -0.47, P < .001) and hip displacement (r = -0.67, P < .001), and the strength of these correlations were comparable to that of DJL. Still, the reason for this interaction was unknown, and muscular strength or muscular activation level might have contributed to this interaction. A further study needs to be done for this interaction in vGRF between ankle brace conditions and landing tasks.

Landing tasks, but not EAS conditions, significantly changed postGRF. Because DJL was more vertical landing tasks; on the other hand, the body mass was moving forward during the FJL. Therefore, to decelerate forward movement of the body, postGRF was greater in FJL compared to DJL. The GRF was correlated with anterior tibiofemoral shear force during a single-leg forward landing task, and the magnitude of the correlation was greater in postGRF compared to vGRF.[184] The anterior tibiofemoral shear force was correlated with postGRF and peak knee extension moment during a stop-vertical jump task.[185,186] This result was not surprising. Therefore, FJL may increase risk of non-contact ACL injury due to higher postGRF and subsequent higher tibiofemoral anterior shear force and knee extension moment. We agreed with a study by Pappas and Carpes[82] that reported drop-landing and forward-landing tasks are different, and the two different landing tasks should be considered as separate tasks. Hence, when we assess the risk of non-contact ACL injury, and landing-mechanics neuromuscular training is implemented, both drop-landing and forward-landing tasks should be practiced.

Limitation

This study had a couple of limitations. Only female subjects who did not participate in intercollegiate athletic sports but were physically active college-age young adults were recruited.

Non-contact ACL injury occurs not only in athletic settings but also any physical activities, so the results of this study can only be generalized to physically active college-age females. Future studies also need to recruit male subjects or different age group subject to confirm the results of this study. Because three dimension motion analysis was not used, the MKD measured in two-dimension kinematic assessment was assumed to represent knee valgus.[45,170] Still, we can not specify the angle of knee valgus in each knee because the MKD simply represent the proportion of the distance between the centers of the patellae relative to the distance between the anterior superior iliac spines. In addition to the assumption of the MKD, we were not able to determine the joint movements in the transverse plane due to using two-dimension kinematic analysis.

Conclusions and Clinical Implication

External ankle supports are often used by health care professionals to reduce the risk of ankle sprains by limiting the frontal plane ankle ROM. However, ankle taping and prophylactic lace up ankle braces sometimes restrict the sagittal plane ankle ROM, as well. The restricted ankle sagittal plane ROM increased peak MKD and postGRF in both drop-jump landing and forward-jump landing tasks. Because excessive knee valgus angle and postGRF stress on the ACL, restricting ankle sagittal plane ROM may also increase the risk of ACL injury due to landing mechanics alteration.

The implication of the study appears complex, especially on vGRF for the two landing tasks. However, it is clear that restricted ankle ROM by prophylactic lace up ankle bracing increased peak MKD postGRF in both drop-jump landing and forward-jump landing tasks. Greater knee valgus angle and postGRF are believed to increase the risk of non-contact ACL injury; therefore, the sagittal plane ankle ROM should not be limited during landing tasks. Because those tasks are frequently performed in sports, health care professionals who apply ankle taping or ankle bracing should be careful not to excessively restrict the sagittal plane ankle ROM. In addition, coaches, strength and conditioning specialists and athletic trainers should be aware of athletes' excessive knee valgus during landing tasks that include landing from a rebound and jump-stop task.

Table 4. Means (*SD*) of Landing Tasks and Ankle Brace Conditions

	DJL			FJL		
	NB	LTB	HTB	NB	LTB	HTB
Ankle Displacement (°)	62.5 (13.2)	51.6 (10.7)	48.2 (10.8)	59.8 (16.6)	50.9 (12.7)	48.0 (12.1)
Knee Displacement (°)	65.7 (11.4)	63.0 (9.9)	60.4 (9.7)	72.4 (12.4)	68.4 (12.6)	65.3 (10.8)
Peak MKD (%)	94 (29.8)	87.1 (33.6)	75.8 (31.2)	96.0 (29.6)	88.7 (31.3)	79.0 (30.3)
vGRF (N/BW)	2.61 (0.43)	2.68 (0.40)	2.76 (0.40)	2.67 (0.36)	2.61 (0.38)	2.51 (0.40)
postGRF (N/BW)	0.35 (0.06)	0.35 (0.05)	0.36 (0.06)	0.63 (0.09)	0.65 (0.07)	0.67 (0.12)

Abbreviations: DJL = Drop-jump landing, FJL = Forward-jump landing, NB = No bracing, LTB = Low-tension bracing, HTB = High-tension bracing, vGRF = vertical ground reaction force, postGRF = posterior ground reaction force.

Table 5. Means (*SD*) of Ankle Range of Motions for Ankle Brace Conditions and Measurement Procedures

	NB			LTB			HTB		
	PF	DF	Total ROM	PF	DF	Total ROM	PF	DF	Total ROM
Knee Extended (°)	44.12 (12.2)	10.42 (5.3)	54.54 (12.1)	33.4 (9.9)	10.1 (4.7)	43.5 (10.9)	22.0 (6.5)	6.8 (6.0)	28.8 (9.9)
Knee Flexed (°)	43.7 (13.6)	16.1 (5.3)	59.7 (14.3)	31.1 (11.0)	14.1 (4.7)	45.2 (11.4)	22.5 (7.3)	11.6 (5.5)	34.1 (9.4)
Lunge DF (°)		43.9 (6.5)			40.7 (4.2)			38.0 (4.3)	

Abbreviations: NB = No bracing, LTB = Low-tension bracing, HTB = High-tension bracing, PF = Plantar flexion, DF = Dorsiflexion, Total ROM = PF + DF

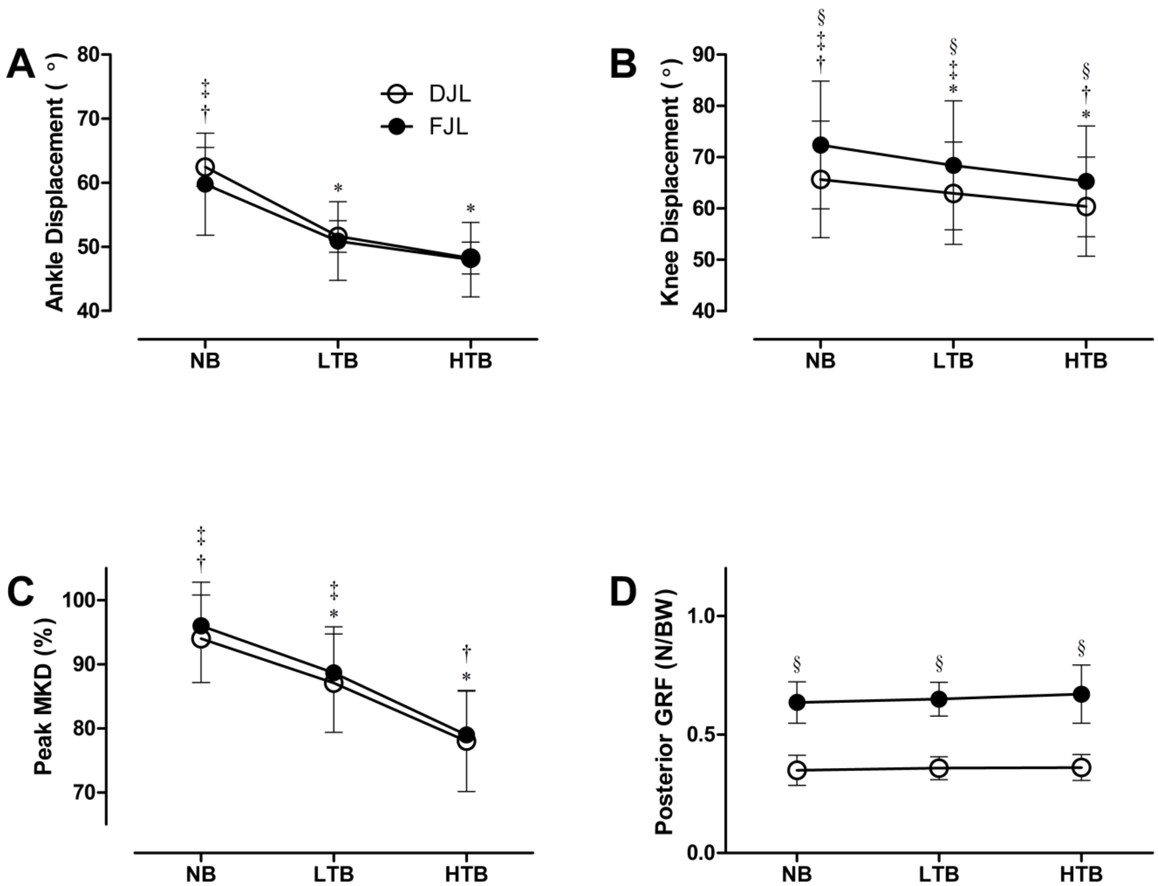

Figure 4. External Ankle Supports and Landing Tasks Affect Landing Kinematics and Kinetics. Ankle displacement (A), knee displacement (B) peak medical knee displacement (peak MKD, C), and posterior ground reaction force (postGRF, D) in three ankle brace conditions (no-bracing = NB, low-tension bracing = LTB, high-tension bracing = HTB) and in two landing tasks (drop-jump landing = DJL, forward-jump landing = FJL). * = significantly differs from NB. † = significantly differs from LTB. ‡ = significantly differs from HTB. § = significantly differs between DJL and FJL tasks.

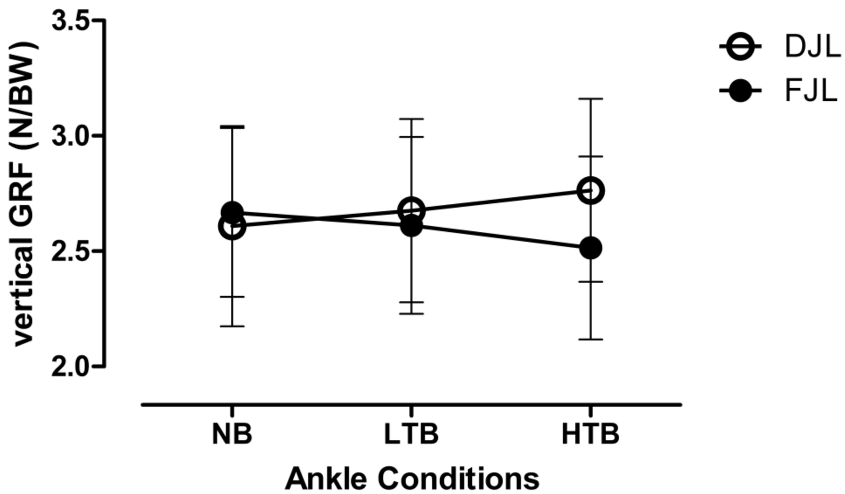

Figure 5. External Ankle Supports Differently Alter Vertical Ground Reaction Force. Vertical ground reaction force (vGRF) in three ankle brace conditions (no-bracing = NB, low-tension bracing = LTB, high-tension bracing = HTB) and in two landing tasks (drop-jump landing = DJL, forward-jump landing = FJL).

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APPENDIX A: DEMOGRAPHIC INFORMATION QUESTIONNAIRE AND ROM

Restricted Ankle Range of Motion Alters Landing Kinematics and Kinetics during Landing Tasks

Demographic Information Sheet

<u>Filled out by Participant</u>:

Age: ______ years old

Dominant leg (Which leg do you use to kick a ball?): LEFT or RIGHT

Injury History:

1. Are you <u>currently</u> injured in the lower extremity (Foot, Ankle, Lower leg, Knee , Thigh, or Hip)? YES NO

2. Do you have any inflammatory signs or symptoms (dysfunction, pain, swelling, heat, redness)? YES NO

3. Have you had surgery in your lower extremity (Foot, Ankle, Lower leg, Knee, Thigh, or Hip) in the <u>last 6 months</u>? YES NO

4. When (DD/ MM/ YYYY) did your last menstrual period end? <u>___/___/ 2013 or 2014</u>

Physical Activity History:

1. What type of physical activity do you usually participate in?

____

2. How many days a week do you usually perform physical activities?

____

3. How long (30 minutes, 45 minutes, 1 hour, etc.) do you usually perform physical activity (in each session)?

____

<u>Filled by CoI</u>

Anthropometric Measurement:

1. Height ____________________cm

2. Weight ____________________kg

Ankle Range of Motion

1. No ankle brace (Baseline)

Knee-extended (Non weight bearing)

Right: Plantar flexion 1)__________ 2)__________

Right: Dorsiflexion 1)__________ 2)__________

Left: Plantar flexion 1)__________ 2)__________

Left: Dorsiflexion 1)__________ 2)__________

Knee-flexed (Non weight bearing)

Right: Plantar flexion 1)__________ 2)__________

Right: Dorsiflexion 1)__________ 2)__________

Left: Plantar flexion 1)__________ 2)__________

Left: Dorsiflexion 1)__________ 2)__________

Lunge dorsiflexion (Weight bearing)

Right: Dorsiflexion 1)__________ 2)__________

Left: Dorsiflexion 1)__________ 2)__________

2. Low tension ankle brace

Knee-extended (Non weight bearing)

Right: Plantar flexion 1)__________ 2)__________

Right: Dorsiflexion 1)__________ 2)__________

Left: Plantar flexion 1)__________ 2)__________

Left: Dorsiflexion 1)__________ 2)__________

Knee-flexed (Non weight bearing)

Right: Plantar flexion 1)__________ 2)__________

Right: Dorsiflexion 1)__________ 2)__________

Left: Plantar flexion 1)__________ 2)__________

Left: Dorsiflexion 1)__________ 2)__________

Lunge dorsiflexion (Weight bearing)

Right: Dorsiflexion 1)__________ 2)__________

Left: Dorsiflexion 1)__________ 2)__________

3. High tension ankle brace

Knee-extended (Non weight bearing)

Right: Plantar flexion 1)__________ 2)__________

Right: Dorsiflexion 1)__________ 2)__________

Left: Plantar flexion 1)__________ 2)__________

Left: Dorsiflexion 1)__________ 2)__________

Knee-flexed (Non weight bearing)

Right: Plantar flexion 1)__________ 2)__________

Right: Dorsiflexion 1)__________ 2)__________

Left: Plantar flexion 1)__________ 2)__________

Left: Dorsiflexion 1)__________ 2)__________

Lunge dorsiflexion (Weight bearing)

Right: Dorsiflexion 1)__________ 2)__________

Left: Dorsiflexion 1)__________ 2)__________

APPENDIX B: INFORMED CONSENT FORM

North Dakota State University
Health, Nutrition, and Exercise Sciences
PO Box 6050 Dept2620
Fargo, ND 58108-6050
701-231-8093

Title of Research Study:
Restricted Ankle Range of Motion Alters Landing Kinematics and Kinetics during Landing Tasks.

This study is being conducted by:
Dr. Pamela Hansen, Associate Professor/ Athletic Training Program Director in the HNES Department (Pamela.J.Hansen@ndsu.edu/ 231-8093), and Hidefusa Okamatsu, a graduate student in the Human Development and Education Wellness option doctoral program (Hidefusa.Okamatsu@ndsu.edu/ 701-388-6825).

Why am I being asked to take part in this research study?
You are being asked to voluntarily participate in this study because you are: (1) healthy and active college female student (between 18 – 25 years old), (2) have no lower extremity muscle or ligament problem, (3) had no surgery in the lower extremities (foot, ankle, lower leg, knee, thigh, or hip) within six months, (4) have no pain or swelling in the lower extremities. If you have any current injury in the lower extremities, or are under age 18 or over 25 years old, you will not be eligible for this study.

What is the reason for doing the study?
This research project examines (1) whether limited ankle movement changes landing technique, and (2) whether vertical and forward landing tasks change landing techniques. Ankle braces or taping are commonly used in athletics; however, it may hinder shock absorbing function of the ankle and produce higher impact at the knee or hip joint. Also, ankle movement is controlled by the calf muscle. Therefore, this study may help people who suffer from knee injuries due to ankle restriction or ankle muscular weakness.

What will I be asked to do?
(1) You will be asked to meet the co-investigator in the Biomechanics Laboratory (BBFH Rm#16) of the BBF. A random number will be assigned to you. You will be asked if you have a history of leg, knee, or ankle injuries, and your last menstrual period. Your height and body weight will be measured. You will be instructed to wear athletic shorts, T-shirts, and low-cut tennis shoes for testing.
(2) You will perform a warm-up for approximately 5 minutes to prepare your body for the landing tasks.
(3) Range of motion of the both your ankles will be measured. Sitting and in standing positions.
(4) Markers will be placed on your shoulders, hips, knees, and ankles. The researcher will demonstrate the jump-landing and forward jump landing tasks for you.

(5) You will be allowed to practice both the drop-jump landing and forward-jump landing tasks until you can perform the sequence properly. *In both tasks, you will be asked to perform a maximum vertical jump immediately after landing.
(6) Following a 2-minute rest, you will be asked to perform the two different landing tasks that will be recorded by digital cameras.

(6a) You will perform drop-jump landing task and forward-jump landing task with no ankle brace, with normal lace-up ankle braces, and tight lace-up ankle braces. Two-minute rest will be given between each task.

(6b) The ankle braces will be fitted on your feet based on your shoe size. Each lace will be tied with a constant tension (normal or tight lace-up brace). The tensions will be measured by a digital tension meter. You will repeat ankle angle measurement and landing tasks.

(7) Once landing tasks are completed, your calf muscle strength will be measured in the Research Laboratory (BBFH Rm#14). You will be sitting and your foot placed in a foot apparatus. You will be instructed to move (push and pull) your foot as fast and hard as you can.
(8) You will perform cool-down exercise (stationary bike) if needed.

<u>Where is the study going to take place, and how long will it take?</u>
You will report to Biomechanics laboratory of the Bentson Bunker Fieldhouse (BBFH Rm#16) on the testing day. The study will last about 60 - 75 minutes.

<u>What are the risks and discomforts?</u>
There are less than minimal risks in the landing tasks because landing and the vertical jump is frequently performed during a physical activity. The vertical jump and strength test with maximum effort may cause muscular soreness and fatigue despite the warm-up session before the testing. The researchers have taken reasonable safeguards to minimize any known risks to you. Exercise testing will be terminated immediately if you feel any pain or discomfort. The researchers are Certified Athletic Trainers who are trained in recognizing and treating exercise-related injuries (e.g., muscle strain, ligamentous sprain, etc.). If needed, EMS will be activated. If you need to visit local outpatient clinics, any costs for medical treatment will be charged to you or your insurance carrier. However, if these costs are not covered by insurance for any reason, they will be your responsibility.

<u>What are the benefits to me?</u>
Your participation to this study will help identify the effect of restricting ankle motion on landing mechanics. This eventually may help your physically active lifestyle with less sports injury. Your participation in this study may also help to identify the effectiveness of ankle braces.

<u>What are the benefits to other people?</u>
The results of this study may contribute to expand the knowledge of ankle braces and knee injuries, especially anterior cruciate ligament injury. The results may help improve other injury prevention protocols, such as anterior cruciate ligament injury prevention exercises.

<u>Do I have to take part in the study?</u>
Your participation is completely voluntary. Your decision whether or not to participate will not affect your relationship with the instructor, or the benefits you may expect to receive from the class and/or NDSU. You are eligible to participate in this study; however, your participation in

this research is completely your choice. Even if you decide to participate in the study, you may change your mind and stop participating at any time without penalty or loss of benefits to which you are already entitled.

Do I receive any compensation for participating in the study?
You will receive $25 cash of compensation for your effort and time commitment. Even if you stop participating, you will still receive $10 no matter how much of the study you complete. The monetary compensation will be provided when you finish or stop the testing session.

What are the alternatives to being in this research study?
Instead of being in this research study, you can choose not to participate.

Who will see the information that I give?
All research records, including the demographic data, ankle motion data, video-recorded landing mechanics, and calf muscle strength will be kept locked storage in the Co-Investigator's office. Your information will be combined with information from other people taking part in the study. When we write about the study, we will write about the combined information that we have gathered. We may publish the results of the study; however, we will keep your name and other identifying information private.

What if I have questions or concerns?
Before you decide whether to accept this invitation to take part in the research study, please ask any questions that might come to mind now. Later, if you have any questions or concerns about the study, you can contact the researchers, Pamela Hansen at 701-231-8093 or Hidefusa Okamatsu at 701-388-6825.

What are my rights as a research participant?
You have rights as a participant in research. If you have questions about your rights, or complaints about this research, you may talk to the researcher or contact the NDSU Human Research Protection Program by:

- Telephone: 701.231.8908
- Email: ndsu.irb@ndsu.edu
- Mail: NDSU HRPP Office, NDSU Dept. 4000, PO Box 6050, Fargo, ND 58108-6050.

The role of the Human Research Protection Program is to see that your rights are protected in this research; more information about your rights can be found at: www.ndsu.edu/research/irb.

Documentation of Informed Consent:
You are freely making a decision whether to be in this research study. Signing this form means that

1. you have read and understood this consent form
2. you have had your questions answered, and
3. you have decided to be in the study.

You will be given a copy of this consent form to keep.

______________________________ ______________

Your signature Date

____

Your printed name

______________________________ ______________

Signature of researcher explaining study Date

____

Printed name of researcher explaining study

APPENDIX C: IRB APPROVAL LETTERS

NDSU NORTH DAKOTA STATE UNIVERSITY

January 17, 2014

FederalWide Assurance FWA00002439

Dr. Pamela Hansen
Department of Health, Nutrition & Exercise Sciences
BBFH 9C

IRB Approval of Protocol #**HE14143**, "**Restricted Ankle Range of Motion Alters Landing Kinetmatics and Kinetics during Landing Tasks**"
Co-investigator(s) and research team: **Hidefusa Okamatsu**

Approval period: 1/17/2014 to 1/16/15 **Continuing Review Report Due: 12/1/14**

Research site(s): **NDSU** Funding agency: **n/a**
Review Type: Expedited category # 4, 6
IRB approval is based on original submission, with revised: protocol and consent form (received 1/17/2014).

Additional approval is required:

- prior to implementation of any proposed changes to the protocol (*Protocol Amendment Request Form*).
- for continuation of the project beyond the approval period (*Continuing Review/Completion Report Form*). A reminder is typically sent two months prior to the expiration date; timely submission of the report is your responsibility. To avoid a lapse in approval, suspension of recruitment, and/or data collection, a report must be received, and the protocol reviewed and approved prior to the expiration date.

A report is required for:

- any research-related injuries, adverse events, or other unanticipated problems involving risks to participants or others within 72 hours of known occurrence (*Report of Unanticipated Problem or Serious Adverse Event Form*).
- any significant new findings that may affect risks to participants.
- closure of the project (*Continuing Review/Completion Report Form*).

Research records are subject to random or directed audits at any time to verify compliance with IRB regulations and NDSU policies.

Thank you for cooperating with NDSU IRB procedures, and best wishes for a successful study.

Sincerely,

Kristy Shirley

Kristy Shirley, CIP
Research Compliance Administrator

INSTITUTIONAL REVIEW BOARD
NDSU Dept 4000 | PO Box 6050 | Fargo ND 58108-6050 | 701.231.8995 | Fax 701.231.8098 | ndsu.edu/irb

Shipping address: Research 1, 1735 NDSU Research Park Drive, Fargo, ND 58102

North Dakota State University
Health, Nutrition, and Exercise Sciences
PO Box 6050 Dept2620
Fargo, ND 58108-6050
701-231-8093

Title of Research Study:
Restricted Ankle range of Motion Alters Landing Kinematics and Kinetics during Landing Tasks.

This study is being conducted by:
Dr. Pamela Hansen, Associate Professor/ Athletic Training Program Director in the HNES Department (Pamela.J.Hansen@ndsu.edu/ 231-8093), and Hidefusa Okamatsu, a graduate student in the Human Development and Education Wellness option doctoral program (Hidefusa.Okamatsu@ndsu.edu/ 701-388-6825).

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What will I be asked to do?
(1) You will be asked to meet the co-investigator in the Biomechanics Laboratory (BBFH Rm#16) of the BBF. A random number will be assigned to you. You will be asked if you have a history of leg, knee, or ankle injuries, and your last menstrual period. Your height and body weight will be measured. You will be instructed to wear athletic shorts, T-shirts, and low-cut tennis shoes for testing.
(2) You will perform a warm-up for approximately 5 minutes to prepare your body for the landing tasks.
(3) Range of motion of the both your ankles will be measured. Sitting and in standing positions.
(4) Markers will be placed on your shoulders, hips, knees, and ankles. The researcher will demonstrate the jump-landing and forward jump landing tasks for you.
(5) You will be allowed to practice both the drop-jump landing and forward-jump landing tasks until you can perform the sequence properly. *In both tasks, you will be asked to perform a maximum vertical jump immediately after landing.
(6) Following a 2-minute rest, you will be asked to perform the two different landing tasks that will be recorded by digital cameras.
(6a) You will perform drop-jump landing task and forward-jump landing task with no ankle brace, with normal lace-up ankle braces, and tight lace-up ankle braces. Two-minute rest will be given between each task.

Institutional Review Board
North Dakota State University
PROTOCOL #: HE14143
APPROVED: 1/17/14
EXPIRES: 1/16/15

(6b) The ankle braces will be fitted on your feet based on your shoe size. Each lace will be tied with a constant tension (normal or tight lace-up brace). The tensions will be measured by a digital tension meter. You will repeat ankle angle measurement and landing tasks.
(7) Once landing tasks are completed, your calf muscle strength will be measured in the Research Laboratory (BBFH Rm#14). You will be sitting and your foot placed in a foot apparatus. You will be instructed to move (push and pull) your foot as fast and hard as you can.
(8) You will perform cool-down exercise (stationary bike) if needed.

<u>Where is the study going to take place, and how long will it take?</u>
You will report to Biomechanics laboratory of the Bentson Bunker Fieldhouse (BBFH Rm#16) on the testing day. The study will last about 60 - 75 minutes.

<u>What are the risks and discomforts?</u>
There are less than minimal risks in the landing tasks because landing and the vertical jump is frequently performed during a physical activity. The vertical jump and strength test with maximum effort may cause muscular soreness and fatigue despite the warm-up session before the testing. The researchers have taken reasonable safeguards to minimize any known risks to you. Exercise testing will be terminated immediately if you feel any pain or discomfort. The researchers are Certified Athletic Trainers who are trained in recognizing and treating exercise-related injuries (e.g., muscle strain, ligamentous sprain, etc.). If needed, EMS will be activated. If you need to visit local outpatient clinics, any costs for medical treatment will be charged to you or your insurance carrier. However, if these costs are not covered by insurance for any reason, they will be your responsibility.

<u>What are the benefits to me?</u>
Your participation to this study will help identify the effect of restricting ankle motion on landing mechanics. This eventually may help your physically active lifestyle with less sports injury. Your participation in this study may also help to identify the effectiveness of ankle braces. However, you may not get any benefit or compensation from being in this research study.

<u>What are the benefits to other people?</u>
The results of this study may contribute to expand the knowledge of ankle braces and knee injuries, especially anterior cruciate ligament injury. The results may help improve other injury prevention protocols, such as anterior cruciate ligament injury prevention exercises.

<u>Do I have to take part in the study?</u>
Your participation is completely voluntary. Your decision whether or not to participate will not affect your relationship with the instructor, or the benefits you may expect to receive from the class and/or NDSU. You are eligible to participate in this study; however, your participation in this research is completely your choice. Even if you decide to participate in the study, you may change your mind and stop participating at any time without penalty or loss of benefits to which you are already entitled.

<u>What are the alternatives to being in this research study?</u>
Instead of being in this research study, you can choose not to participate.

<u>Who will see the information that I give?</u>
All research records, including the demographic data, ankle motion data, video-recorded landing mechanics, and calf muscle strength will be kept locked storage in the Co-Investigator's office. Your information will be combined with information from other people taking part in the study. When we write about the study, we will write about the combined information that we have gathered. We may

publish the results of the study; however, we will keep your name and other identifying information private.

What if I have questions or concerns?

Before you decide whether to accept this invitation to take part in the research study, please ask any questions that might come to mind now. Later, if you have any questions or concerns about the study, you can contact the researchers, Pamela Hansen at 701-231-8093 or Hidefusa Okamatsu at 701-388-6825.

What are my rights as a research participant?

You have rights as a participant in research. If you have questions about your rights, or complaints about this research, you may talk to the researcher or contact the NDSU Human Research Protection Program by:

- Telephone: 701.231.8908
- Email: ndsu.irb@ndsu.edu
- Mail: NDSU HRPP Office, NDSU Dept. 4000, PO Box 6050, Fargo, ND 58108-6050.

The role of the Human Research Protection Program is to see that your rights are protected in this research; more information about your rights can be found at: www.ndsu.edu/research/irb.

Documentation of Informed Consent:

You are freely making a decision whether to be in this research study. Signing this form means that

1. you have read and understood this consent form
2. you have had your questions answered, and
3. you have decided to be in the study.

You will be given a copy of this consent form to keep.

________________________________ ____________

Your signature Date

____

Your printed name

________________________________ ____________

Signature of researcher explaining study Date

____

Printed name of researcher explaining study

NDSU NORTH DAKOTA STATE UNIVERSITY
INSTITUTIONAL REVIEW BOARD
office: Research 1, 1735 NDSU Research Park Drive, Fargo, ND 58102
mail: NDSU Dept. #4000, PO Box 6050, Fargo, ND 58108-6050
p: 701.231.8995 f: 701.231.8098 e: ndsu.irb@ndsu.edu w: www.ndsu.edu/irb

Date Received
RECEIVED
JAN 23 2014
INSTITUTIONAL REVIEW BOARD

IRB Protocol #: HE14143

Protocol Amendment Request Form

Changes to approved research may not be initiated without prior IRB review and approval, except where necessary to eliminate apparent immediate hazards to participants. Reference: SOP 7.5 Protocol Amendments.

Examples of changes requiring IRB review include, but are not limited to changes in: investigators or research team members, purpose/scope of research, recruitment procedures, compensation strategy, participant population, research setting, interventions involving participants, data collection procedures, or surveys, measures or other data forms.

Protocol Information:

Range
Protocol #: **HE14143** Title: **Restricted Ankle Rnage of Motion Alters Landing kinematics and Kinetics during Landing Tasks**

Review category: ☐ Exempt ☒ Expedited ☐ Full board

Principal investigator: **Dr. Pamela Hansen** Email address: **Pamela.J.Hansen@ndsu.edu**
Dept: **HNES**

Co-investigator: **Hidefusa Okamatsu** Email address: **Hidefusa.Okamatsu@ndsu.edu**
Dept: **Human Development and Education**

Principal investigator signature, Date: Pamela Hansen (email) 1/23/14

In lieu of a written signature, submission via the Principal Investigator's NDSU email constitutes an acceptable electronic signature.

Description of proposed changes:

1. Date of proposed implementation of change(s)*: **1/24/2014**
   ** Cannot be implemented prior to IRB approval unless the IRB Chair has determined that the change is necessary to eliminate apparent immediate hazards to participants.*

2. Describe proposed change(s), including justification:
   Original IRB protocol did not include financial compensation for participating in the study. In this amendment, a change has been made to the (4) Compensation in the Recruitment section of the IRB protocol form. According to this change, the section of "What are the benefits to me?" in the Consent form has been revised, as well.
   Each participant will receive $25 for completing the study. Even if the subject changes her mind and

decides to stop participating in the middle of the study, she will receive $10 for her effort and time commitment.

3. Will the change involve a change in principal or co- investigator?

   ☒ No - skip to Question 4

   ☐ Yes:

   - *Include an Investigator's Assurance (last page of protocol form), signed by the new PI or co-investigator*
   - *Conflict of Interest disclosure.* Does any investigator responsible for the design, conduct or reporting of the project (including their immediate family members) have a financial, personal or political interest that may conflict with their responsibility for protecting human participants in NDSU research? (*SOP 6.2 Conflict of Interest in Human Research, Investigator and Research Team)*

     ☐ No – As PI, I attest that I have conferred with my co-investigators and key personnel and confirmed that no financial, personal or political interests currently exist related to this research.

     ☐ Yes – Describe the related financial, personal or political interests, and **attach documentation of COI disclosure and review** *(as applicable).*

     Financial, personal or political interests related to the research (the sponsor, product or service being tested, or a competing product or service) may include:

     - compensation (e.g., salary, payment for services, consulting fees)
     - intellectual property rights or equity interests
     - board memberships or executive positions
     - enrollment or recruitment bonus payments

     (Refer to *NDSU Policy 151.1, External Activities and Conflicts of Interest, and NDSU Policy 823, Financial Disclosure – Sponsored Projects* for specific disclosure requirements.)

Note: If the change is limited to addition/change in research team members, skip the rest of this form.

4. Will the change(s) increase any risks, or present new risks *(physical, economic, psychological, or sociological)* to participants?

   ☒ No

   ☐ Yes: *In the appropriate section of the protocol form, describe new or altered risks and how they will be minimized.*

5. Does the proposed change involve the addition of a vulnerable group of participants?

   Children: ☒ no ☐ yes – include the *Children in Research* attachment form

   Prisoners: ☒ no ☐ yes – include the *Prisoners in Research* attachment form

   Cognitively impaired individuals: ☒ no ☐ yes*

   Economically or educationally disadvantaged individuals: ☒ no ☐ yes*

   **Provide additional information where applicable in the revised protocol form.*

6. Does the proposed change involve a request to waive some or all the elements of informed consent or documentation of consent?

   ☒ no

☐ yes - Attach the *Informed Consent Waiver or Alteration Request.*

7. Does the proposed change involve a new research site?
   ☒ no
   ☐ yes

If information in your previously approved protocol has changed, or additional information is being added, incorporate the changes into relevant section(s) of the protocol. Highlight (e.g. print and highlight the hard copy, or indicate changes using all caps, asterisks, etc) the changed section(s) and attach a copy of the revised protocol to this form. *(If the changes are limited to addition/change in research team members, a revised protocol form is not needed.)*

Impact for Participants (future, current, or prior):

1. Will the change(s) alter information on previously approved versions of the recruitment materials, informed consent, or other documents, or require new documents?
   ☒ No
   ☐ Yes - attach revised/new document(s)

2. Could the change(s) affect the willingness of *currently* enrolled participants to continue in the research?
   ☒ No
   ☐ Yes - describe procedures that will be used to inform current participants, and re-consent, if necessary:

3. Will the change(s) have any impact to *previously* enrolled participants?
   ☒ No
   ☐ Yes - describe impact, and any procedures that will be taken to protect the rights and welfare of participants:

----------FOR IRB OFFICE USE ONLY--------

Request is: ☒ **Approved** ☐ **Not Approved**	
Review: ☐**Exempt, category#:** ____ ☐**Expedited method, category #** ____ ☐**Convened meeting, date:** ____ ☒**Expedited review of minor change**	
IRB Signature: Kristy Shirley	**Date:** 1/27/14

Comments:

North Dakota State University
Health, Nutrition, and Exercise Sciences
PO Box 6050 Dept2620
Fargo, ND 58108-6050
701-231-8093

Title of Research Study:
Restricted Ankle Range of Motion Alters Landing Kinematics and Kinetics during Landing Tasks.

This study is being conducted by:
Dr. Pamela Hansen, Associate Professor/ Athletic Training Program Director in the HNES Department (Pamela.J.Hansen@ndsu.edu/ 231-8093), and Hidefusa Okamatsu, a graduate student in the Human Development and Education Wellness option doctoral program (Hidefusa.Okamatsu@ndsu.edu/ 701-388-6825).

Why am I being asked to take part in this research study?
You are being asked to voluntarily participate in this study because you are: (1) healthy and active college female student (between 18 – 25 years old), (2) have no lower extremity muscle or ligament problem, (3) had no surgery in the lower extremities (foot, ankle, lower leg, knee, thigh, or hip) within six months, (4) have no pain or swelling in the lower extremities. If you have any current injury in the lower extremities, or are under age 18 or over 25 years old, you will not be eligible for this study.

What is the reason for doing the study?
This research project examines (1) whether limited ankle movement changes landing technique, and (2) whether vertical and forward landing tasks change landing techniques. Ankle braces or taping are commonly used in athletics; however, it may hinder shock absorbing function of the ankle and produce higher impact at the knee or hip joint. Also, ankle movement is controlled by the calf muscle. Therefore, this study may help people who suffer from knee injuries due to ankle restriction or ankle muscular weakness.

What will I be asked to do?
(1) You will be asked to meet the co-investigator in the Biomechanics Laboratory (BBFH Rm#16) of the BBF. A random number will be assigned to you. You will be asked if you have a history of leg, knee, or ankle injuries, and your last menstrual period. Your height and body weight will be measured. You will be instructed to wear athletic shorts, T-shirts, and low-cut tennis shoes for testing.
(2) You will perform a warm-up for approximately 5 minutes to prepare your body for the landing tasks.
(3) Range of motion of the both your ankles will be measured. Sitting and in standing positions.
(4) Markers will be placed on your shoulders, hips, knees, and ankles. The researcher will demonstrate the jump-landing and forward jump landing tasks for you.
(5) You will be allowed to practice both the drop-jump landing and forward-jump landing tasks until you can perform the sequence properly. *In both tasks, you will be asked to perform a maximum vertical jump immediately after landing.
(6) Following a 2-minute rest, you will be asked to perform the two different landing tasks that will be recorded by digital cameras.

(6a) You will perform drop-jump landing task and forward-jump landing task with no ankle brace, with normal lace-up ankle braces, and tight lace-up ankle braces. Two-minute rest will be given between each task.

Institutional Review Board
North Dakota State University
PROTOCOL #: HE14143
APPROVED: 1/28/14
EXPIRES: 1/16/15

(6b) The ankle braces will be fitted on your feet based on your shoe size. Each lace will be tied with a constant tension (normal or tight lace-up brace). The tensions will be measured by a digital tension meter. You will repeat ankle angle measurement and landing tasks.
(7) Once landing tasks are completed, your calf muscle strength will be measured in the Research Laboratory (BBFH Rm#14). You will be sitting and your foot placed in a foot apparatus. You will be instructed to move (push and pull) your foot as fast and hard as you can.
(8) You will perform cool-down exercise (stationary bike) if needed.

<u>Where is the study going to take place, and how long will it take?</u>
You will report to Biomechanics laboratory of the Bentson Bunker Fieldhouse (BBFH Rm#16) on the testing day. The study will last about 60 - 75 minutes.

<u>What are the risks and discomforts?</u>
There are less than minimal risks in the landing tasks because landing and the vertical jump is frequently performed during a physical activity. The vertical jump and strength test with maximum effort may cause muscular soreness and fatigue despite the warm-up session before the testing. The researchers have taken reasonable safeguards to minimize any known risks to you. Exercise testing will be terminated immediately if you feel any pain or discomfort. The researchers are Certified Athletic Trainers who are trained in recognizing and treating exercise-related injuries (e.g., muscle strain, ligamentous sprain, etc.). If needed, EMS will be activated. If you need to visit local outpatient clinics, any costs for medical treatment will be charged to you or your insurance carrier. However, if these costs are not covered by insurance for any reason, they will be your responsibility.

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Your participation to this study will help identify the effect of restricting ankle motion on landing mechanics. This eventually may help your physically active lifestyle with less sports injury. Your participation in this study may also help to identify the effectiveness of ankle braces.

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<u>Do I receive any compensation for participating in the study?</u>
You will receive $25 cash of compensation for your effort and time commitment. Even if you stop participating, you will still receive $10 no matter how much of the study you complete. The monetary compensation will be provided when you finish or stop the testing session.

<u>What are the alternatives to being in this research study?</u>
Instead of being in this research study, you can choose not to participate.

<u>Who will see the information that I give?</u>

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________________________________ ____________

Your signature Date

____

Your printed name

________________________________ ____________

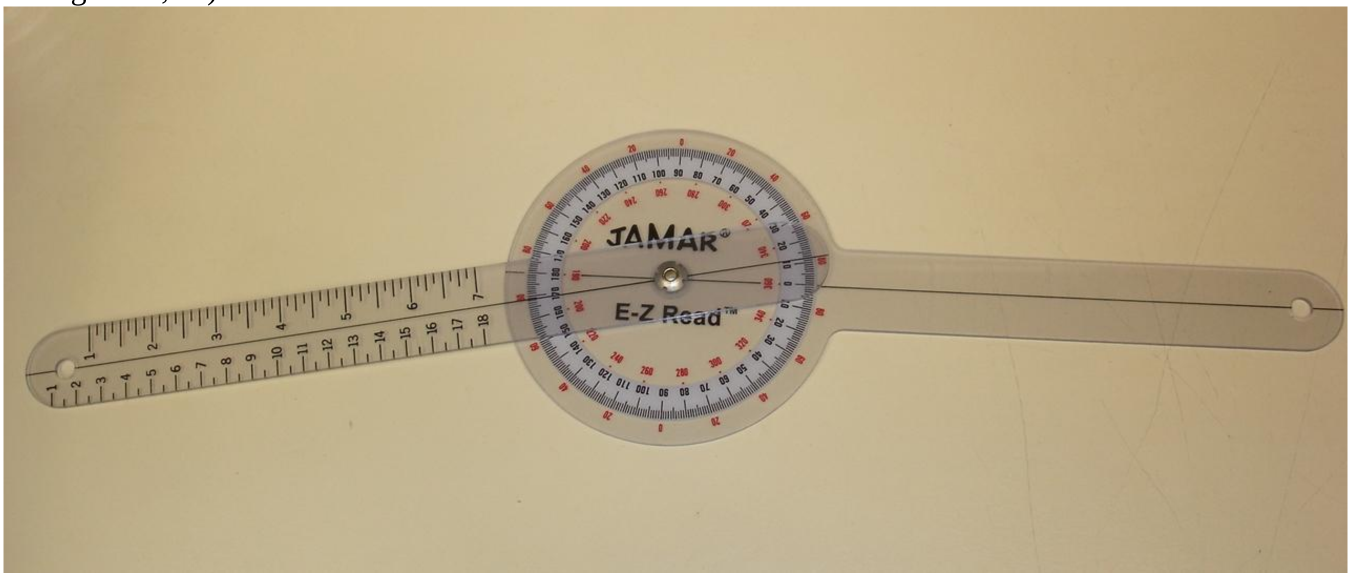
Signature of researcher explaining study Date

____

Printed name of researcher explaining study

APPENDIX D: MATERIAL USED

1-degree-increment transparent plastic goniometer (Patterson Medical/ Sammons Preston, Bolingbrook, IL)



A hand-held digital scale (Rapala VMC, Vaaksy, Finland)

Each lace will be held at the target tension for one second at 90° angle from the surface of the ankle brace.