Landing Error Scoring System Evaluation of Anterior Cruciate Ligament Injury Risk with Prophylactic Ankle Support

A thesis presented to
the faculty of
the College of Health Sciences and Professions of Ohio University

In partial fulfillment
of the requirements for the degree
Master of Science

Marseille A. Mosher
May 2015

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This thesis titled
Landing Error Scoring System Evaluation of Anterior Cruciate Ligament Injury Risk
with Prophylactic Ankle Support

by
MARSEILLE A. MOSHER

has been approved for
the School of Applied Health Sciences and Wellness
and the College of Health Sciences and Professions by

Jeffrey A. Russell
Assistant Professor of Applied Health Sciences and Wellness

Randy Leite
Dean, College of Health Sciences and Professions
Abstract

MOSHER, MARSEILLE A., M.S., May 2015, Athletic Training

Landing Error Scoring System Evaluation of Anterior Cruciate Ligament Injury Risk with Prophylactic Ankle Support

Director of Thesis: Jeffrey A. Russell

Background: Lateral ankle sprains are prevented through the use of external ankle support including taping or bracing. Both types of support restrict ankle dorsiflexion, which may cause biomechanical alterations at the knee. The Landing Error Scoring System (LESS) measures the number of biomechanical errors committed during a jump landing to determine risk of anterior cruciate ligament (ACL) injury. Hypothesis: Ankle taping and bracing will decrease passive ankle dorsiflexion stiffness (PADS), thus increasing the risk of ACL injury. Methods: Participants wore tape, brace, and no external ankle support for three conditions. Each completed LESS and had PADS measured with the modified Lidcombe template. Results: Wearing the brace caused more errors on the LESS; however, neither the tape nor brace had an effect on PADS. Conclusion: The results of this study provide insight for the decision to prevent lateral ankle sprains using prophylactic braces at the risk of jeopardizing the ACL.
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Chapter 1: Introduction

Lateral ankle sprains are one of the most common musculoskeletal injuries with an incidence rate much higher than anterior cruciate ligament (ACL) tears. A meta-analysis in 2013 found the rate to be as high as 13.6 per 1,000 exposures for females and 6.94 for males.\(^1\) Depending on its severity, an ankle sprain can result in high medical costs, extended time loss from activity, and likely recurrence of injury. Therefore, a method either to prevent initial sprains or to allow return to play sooner following an ankle injury would be beneficial. A more conservative approach involves rehabilitation; however, most patients and clinicians opt for some form of external ankle support that prevents the inversion mechanism of injury such as modified footwear, ankle taping, or braces.\(^2\)

Ankle inversion is not the only motion that is affected by external support. Lace-up ankle braces and taping techniques have an influence on sagittal plane motion by decreasing the maximum range of ankle dorsiflexion (DF).\(^3\)–\(^5\) Theoretically, if ankle DF is decreased, then the amount of vertical ground reaction force (GRF) transmitted to the knee should be higher.\(^6\) This increased force creates shearing within the knee joint, which, coupled with biomechanical risk factors such as knee valgus or decreased knee flexion, can potentially cause injury to the main stabilizing ligament of the knee, the ACL.

Rupture of the ACL results in serious outcomes for an athlete due to its lengthy recovery time and cost of surgical repair. Although many athletes choose to undergo reconstructive surgery, there are still long-term complications that lead to early
development of osteoarthritis within the knee joint.\textsuperscript{7–9} The prevalence of ACL injuries and surgical reconstructions remains relatively low among all levels of competition and sports. High school athletes have an incidence rate of 0.061 injuries per 1,000 exposures whereas collegiate athletes have a rate of 0.150.\textsuperscript{10} Comparatively, college football produces an incidence rate of 0.142 per 1,000 exposures, with the majority of them occurring during games and postseason competition when the knee is more fatigued.\textsuperscript{11} Females have an incidence rate twice that of males because they exhibit a greater preponderance of associated risk factors.\textsuperscript{8,11–18}

Risk factors for ACL injury can be extrinsic or intrinsic, including environmental, anatomical, neuromuscular, and biomechanical. These factors must be identified in order to prevent the non-contact mechanism of injury that accounts for nearly 70\% of all ACL injuries.\textsuperscript{19–21} A non-contact injury occurs as a result of a jump landing or sudden reaction to a situation during athletic play. The mechanism can involve a number of adverse joint motions including trunk rotation, hip adduction, knee valgus and hyperextension, and tibial internal rotation.\textsuperscript{14}

Athletes who experience these potentially harmful motions consistently during activity should be introduced to an ACL prevention program that focuses on strengthening the muscles that counteract those movements. Poor biomechanics, along with neuromuscular risk factors, can be improved in order to minimize the risk of injury. Some factors cannot be altered, including environmental conditions, joint laxity, intercondylar notch width, and sex.\textsuperscript{19,22} Individuals with these risk factors should be monitored closely for potential injury and included in an injury prevention program in
order to establish and maintain proper biomechanics and neuromuscular control during activity.

The complexity, cost, and time necessary to collect data to evaluate biomechanics does not benefit clinicians as much as having an easy, fast, and comprehensive screening during preseason to identify athletes that are at risk for an ACL injury. Because of this, the Landing Error Scoring System (LESS)\(^\text{18}\) was developed as a helpful tool for clinicians. Three-dimensional (3-D) motion analysis is still considered the gold standard, but the LESS can instead be used to identify those who are at an increased risk.\(^\text{18,21}\) Studies identifying ACL injury risk factors have successfully utilized 3-D motion analysis;\(^\text{13,23,24}\) however, in comparison to this type of measurement, the LESS exhibits concurrent and discriminant validity, as well as good reliability, making it useful for clinical application.\(^\text{18,21}\)

While the LESS can be used to determine whether or not an athlete has a high or low risk of ACL injury based on the number of errors committed during a jump landing, few studies have analyzed how the LESS can detect changes in that level of risk when a certain biomechanical factor is manipulated. Ankle dorsiflexion is one of the first motions to occur during dynamic closed-kinetic chain activity, particularly a jump landing. It is one of the most essential kinematic motions because it dissipates the ground reaction force throughout the entire lower extremity kinetic chain.\(^\text{6}\) A few studies\(^\text{25–27}\) investigated how changes in the amount of DF can influence biomechanics at the knee, but none determined what influence this change has on ACL injury risk.
Prior to investigating how external ankle support, such as tape and bracing, affect ACL injury risk, it is important to determine that both the tape and brace have an effect on joint compliance during a jump landing. Individuals with less ankle DF ROM often have limitations due to soft tissue restrictions of the posterior structures of the leg. A calcaneal tendon or Achilles that is not very compliant transfers more energy to the proximal joints. Achilles tendon compliance can be observed through goniometry, or more accurately, by measuring passive joint stiffness.\textsuperscript{28–30} The Achilles is not the only tendon that causes dorsiflexion stiffness, but it is the largest contributor compared to those of the deep posterior compartment.

Since ankle braces and taping decrease ankle DF in a way similar to the effect of a relatively taut Achilles tendon, it is possible that increased joint stiffness will be present when external support is worn. Measurement of this joint stiffness may further validate any significant results related to a change in ACL injury risk when wearing external ankle support because it can determine that the brace and tape are contributing factors instead of just an individual’s soft tissue limitations.

In light of current evidence, the purpose of this study is to determine if a decrease in ankle DF caused by wearing an ankle brace or tape increases the risk of ACL injury as determined by the LESS. This within-subjects repeated measures study in a controlled laboratory setting evaluated how an individual’s passive ankle dorsiflexion stiffness and jump-landing biomechanics change when wearing an ankle tape or brace when compared to a control. The ASO ankle brace (Medical Specialties, Charlotte, NC) was used because it is the most commonly recommended brace by clinicians.\textsuperscript{31} The basket weave taping
technique was used for this study since it is the most commonly used method in athletic training programs. Finding a relationship between ankle DF and the overall LESS score would allow athletes, coaches, and healthcare providers to make evidenced-based decisions about whether or not using external ankle support is advisable in light of its potential for increasing the risk of ACL injury.

**Research Questions and Hypotheses**

1. Does wearing external ankle support decrease the amount of passive ankle dorsiflexion available for weight bearing activity?
   a. The ankle brace and tape will limit motion due to increased passive stiffness.
2. Does wearing external ankle support increase the number of biomechanical errors during a jump landing?
   a. Wearing an ankle brace or tape will result in a higher overall score on the LESS, indicating an increased risk of ACL injury.
3. Is there a difference between the lace-up ASO brace and the basket weave taping technique on ankle dorsiflexion stiffness and jump-landing biomechanics?
   a. Wearing the ASO ankle brace will result in a greater amount of passive stiffness and higher overall score on the LESS.
Independent Variables

1. Condition
   a. No external ankle support (control).
   b. Basket weave taping technique.
   c. Lace-up ASO brace.

Dependent Variables

1. Passive ankle dorsiflexion stiffness coefficient.
2. Overall LESS scores.

Assumptions

1. The tape was applied without much variation for each participant.
2. Participants did not knowingly alter their jump-landing technique because of
   the brace or tape.
3. Evaluation of the video data was consistent for each participant.

Limitations

1. Using only healthy participants with no previous history of ACL injury
   prevents the ability to determine any change in risk for reinjury for those that
   have had ACL reconstruction surgery.
2. The results of this study cannot be generalized to recreational, high school, or
   professional athletes.
3. The time allotted only allows single episode measures of external ankle
   support use instead of a prolonged period of physical activity with this type of
   intervention.
Delimitations

1. Participants were students at Ohio University who were healthy and physically active members of their respective sports team.

2. Measurements were obtained during a single testing session rather than requiring participants to report on three separate occasions to complete trials for each condition.

3. The evaluator of the LESS videos for this study was also the main researcher and could be biased due to not being blinded.

4. Measurements for the modified Lidcombe template were taken in a position such that the knee was extended in order to account for the influence of the gastrocnemius. No measurements were taken with the knee flexed.

5. The modified Lidcombe template was used to measure passive ankle dorsiflexion stiffness even though face validity was assumed.
Chapter 2: Review of the Literature

Introduction

Sports that require high-risk skills such as cutting, jumping, sudden deceleration, and contact with other players have a higher chance of exposure to forces that affect the integrity of the anterior cruciate ligament (ACL). These sports include football, soccer, basketball, and lacrosse. Noncontact ACL injuries in these sports often occur while avoiding contact or reacting suddenly to a play; however, the most common mechanism of injury is a jump landing where the knee is placed in a hyperextended and valgus position. Therefore, the best method for measuring a person’s susceptibility to a non-contact injury is through biomechanical analysis of one’s jump-landing technique.

Three-dimensional (3-D) motion analysis is considered the most accurate tool for measuring kinematics. Unfortunately, 3-D motion analysis is expensive, time consuming, and impractical for clinical use when conducting large preseason screenings. The Landing Error Scoring System (LESS) was developed as a more practical and feasible tool to determine ACL injury risk based on the number of errors committed when completing a jump landing. LESS overall scores provide concurrent and discriminately valid results and has good interrater and intrarater reliability. One study did not find the LESS to have good predictive value for ACL injury; yet, the test is still considered useful for identifying risk.

One potential biomechanical influence that has yet to be evaluated is whether or not external ankle support, such as ankle taping or bracing, affects the likelihood of
sustaining an ACL injury. Dorsiflexion is essential to dissipating ground reaction forces (GRFs) and maintaining balance during a jump landing, so if this motion is restricted by external support, then the GRFs will be attenuated to the knee and affect the ACL.\textsuperscript{6,7,24–26} Although prior studies\textsuperscript{25,26,34} have evaluated the change in knee biomechanics when wearing a prophylactic ankle brace or taping, it is unknown how the risk of ACL injury is directly affected by these interventions. If the risk of an ACL injury, as determined by the LESS, increases with the application of a brace or tape, then trying to protect the ankle may put the knee at a greater risk for injury. Therefore, the purpose of this study is determine whether or not using an ankle brace or tape increases the risk of ACL injury as determined by the LESS.

Risk Factors for ACL Injury

**External factors.** Studies of external environmental factors that can cause ACL injury are limited due to the difficulties of isolating each factor. ACL injuries are more frequent during high-evaporation and low-rainfall periods that result in a harder grass surface, which allows more traction between the shoe and the ground.\textsuperscript{22} Surfaces such as artificial turf that have a high degree of friction and resistance to movement compared to soft grass, allow for a greater chance that the foot ends up in a more fixed position.\textsuperscript{19} As many as 80% of non-contact injuries occur when the foot is in a fixed position since the tibia can internally rotate and the body weight shifts medially over the knee, stressing the ACL. Although playing surface may be a considerable contributing factor to risk of injury, footwear also plays a substantial role in the shoe-surface interaction. The difficulty in assessing the associated environmental risk based on shoe-surface interaction
is that there are various combinations of footwear and playing surfaces that an athlete may encounter.

**Anatomical factors.** The mechanical alignment of the lower kinetic chain can be altered considerably with an anatomical abnormality since each joint influences the rest of the lower extremity during normal function. Proper alignment allows optimal biomechanical and neuromuscular performance during gait, as well as joint stability, particularly at the knee. The Q-angle, pelvic posture, knee joint laxity, and intercondylar notch width are all anatomical properties associated with increased risk of ACL injury. Unfortunately, risk related to these factors is not easily mitigated because one’s anatomy cannot be modified. The recognition of these risk factors is beneficial for the identification of those who are most likely to sustain an ACL injury in order for a prevention program to be introduced.

The Q-angle is formed by two lines intersecting at the central patella. One line is directed from the patella to the anterior superior iliac spine of the pelvis, while the other is directed toward the tibial tuberosity. Due to women having a greater pelvic width relative to femur length, women have a relatively higher Q-angle compared to males; this is one of the many reasons why females sustain more ACL injuries. A larger Q-angle increases the static and dynamic stresses applied to the knee because it creates a valgus alignment. Valgus alignment can be a predictor of a valgus knee position during a jump landing; therefore, the static Q-angle measurement is a good predictor of dynamic knee positions. Shambaugh, Klein, and Herbert reported a higher mean Q-angle in subjects sustaining an ACL injury. This is why valgus alignment, described as medial knee
position greater than the midfoot at initial contact, is considered to be an error on the LESS.

Valgus knee position is not the only accessory movement that is correlated with a larger Q-angle. One study found that when compared to being in subtalar neutral, women who had a greater amount of pronation in a navicular drop test had a greater change in Q-angle. Wearing an ankle brace can decrease the amount of supination and tibial external rotation that occurs during a jump landing. Therefore, if someone exhibits hyperpronation with a large Q-angle, an ankle brace could adversely affect motion at the knee during dynamic activity.

Another anatomical abnormality that can be associated with increased knee valgus is found in the proximal portion of the lower kinetic chain at the pelvis. Anterior pelvic tilt places the hip in an internally rotated and flexed position. This posture alters the moment arms of the hamstrings and gluteal muscles, which influences motion at the knee. The gluteus medius performs hip abduction and prevents knee adduction, or valgus. The hamstrings prevent anterior tibial translation, which is the arthokinematic motion that causes ACL injury due to hyperextension and valgus stress. During gait, the indirect effects of an anterior pelvic tilt is evident at the distal portion of the lower kinetic chain. A valgus knee alignment due to poor gluteal strength causes internal tibial rotation, which is associated with excessive pronation at the subtalar joint. Excessive pronation is commonly found in ACL-injured patients because of the stress on the ACL from the internal tibial rotation. This dilemma will only be exacerbated when an ankle brace is used because it limits the amount of external tibial rotation that occurs.
Anatomical factors that affect the ACL directly within the tibiofemoral joint are increased laxity and intercondylar notch width. Generalized joint laxity leads to genu recurvatum and excessive hyperextension during activity. Anterior-posterior joint laxity is related to an increased risk of ACL injury because the translation of the tibia on the femur stresses the fibers of the ACL in the sagittal plane. This risk factor is found more often in females than males, most likely due to hormonal influences. Estrogen affects the tensile properties of ligaments; when its systemic level surges during the pre-ovulatory phase of the menstrual cycle, the ligaments of the body are more susceptible to injury. Sex differences have also been associated with structural characteristics of the ACL such as length and cross-sectional area (CSA). Chandrashekar reported that women’s ACLs are shorter, have a smaller CSA, and have a lower fibril concentration to create tensile strength than those seen in men. These characteristics decrease the threshold for the point of failure, making the ligament more susceptible to injury with an applied stress.

A recent study also found a significant linear relationship between ACL CSA and size of the intercondylar notch. Although the direct role that a small intercondylar notch plays in the risk of injuring the ACL is unclear, it is speculated that a notch that is smaller than typical allows for the ACL to be impinged during external tibial rotation. Thus, a weak ACL based on small CSA may be what predisposes someone to injury, but a small notch width may further add to insult.

All of these anatomical risk factors are only one small part of an evaluation that can determine whether or not someone is susceptible to an ACL injury. Knowledge of
these risk factors leads to a better understanding about how the ACL functions and is injured, but it is important to realize that they cannot be altered in order to decrease the associated risk. Anyone presenting with one or more of these anatomical risk factors may need to be cognizant if wearing external ankle support because it can further exacerbate one’s anatomical variations, such as a larger Q-angle or anterior pelvic tilt.

**Neuromuscular factors.** While dynamic stabilizers are essential to protecting a joint, they are also important for protecting the static stabilizers within the joint. The hamstrings secondarily protect the ACL because they prevent anterior translation of the tibia. If the hamstrings have adequate strength and stiffness, they can serve as a primary restraint when an anterior force is applied, thus reducing strain on the ACL. Therefore, poor hamstring strength and stiffness along with improper co-activation with the quadriceps may increase the risk of ACL injury.

The hamstrings and quadriceps form an agonist-antagonistic relationship about the knee joint that is crucial for optimal joint stability. Contractions of the quadriceps place stress on the ACL when the knee is flexed 30°. Eccentric hamstring strength is crucial to reduce the load on the ACL in order to prevent anterior tibial translation when the quadriceps extend the knee. Fortunately, the hamstrings are at a structural advantage to the quadriceps because they have three musculotendinous attachments compared to the one for the quadriceps on the tibial tuberosity. While the hamstrings function primarily in the sagittal plane, they each perform additional motions within the frontal and transverse planes that are crucial to protecting the integrity of the ACL.
The semitendinosus and semimembranosus muscles of the hamstring group insert on the medial aspect of the tibia, thus producing internal rotation and adduction motions at the knee. On the other hand, the biceps femoris inserts on the lateral aspect of the tibia and fibular head, producing the antagonistic motions of external rotation and abduction. When the biceps femoris contracts simultaneously with the tensor fascia lata, they together help prevent excessive internal tibial rotation, which is the main stress on the ACL within the transverse plane. Because an ACL tear is associated with altered movement patterns in multiple planes, it is crucial to have three-dimensional strength and neuromuscular control of the hamstrings.

Individuals that perform activity with a quadriceps-dominant contraction during dynamic activities such as cutting or jump landing cause excessive strain on the ACL because the hamstrings are not strong enough eccentrically to counteract the concentric quadriceps contraction. Myer et al. found the ratio of quadriceps to hamstring strength to be significantly different between males and females. The females’ quadriceps strength was comparable to the males; however, the hamstring strength was considerably lower. Poor neuromuscular control of the hamstrings exacerbates any biomechanical factors associated with a jump landing, thus increasing the risk of an ACL injury even further.

Muscular stiffness is also an important factor in neuromuscular control. Wearing external ankle support can potentially increase the amount of stiffness at the ankle and the amount of dorsiflexion range of motion available during a jump landing. This change affects the rest of the lower extremity, particularly the hamstrings, because they are a part of the posterior kinetic chain. Hamstrings that display a large amount of stiffness produce
increased knee flexion and a decreased internal knee-varus moment. However, if flexion ROM is altered by limitations at the ankle, knee joint stiffness can be affected. Poor hamstring stiffness adversely affects the knee because the hamstrings decrease anterior tibial shearing forces through arthrokinematic motion in the sagittal plane. Hamstring stiffness and ROM play a significant role in protecting the ACL during dynamic activity.

Neuromuscular control of the hamstrings has a direct effect on protecting the knee, but core stability has an equally important indirect effect. Having adequate core stability allows one to maintain position of one’s body following any sort of perturbation. Poor neuromuscular control of the trunk allows uncontrolled trunk movement during activity, changing the position of the center of mass in relation to the knee joint. This can cause injury to any part of the lower extremity including the knee and the ankle. Those who suffer from lateral ankle sprains have greater postural sway and poor trunk proprioception, meaning that if athletes with a previous history of ankle sprains are wearing an external ankle support, they already exhibit one of the risk factors that could further exacerbate their risk of ACL injury.

**Biomechanics of ACL Pathology**

Although static measurements such as Q-angle and postural alignments are helpful in determining the risk of ACL pathology, most injuries occur while performing a dynamic task. Biomechanical analysis of the lower extremity provides better insight into how an individual’s joints move in space during activity. The kinematics and kinetics indicate the degree of deviation from normal joint alignments as well as the amount of
torque applied to the joint. Because most ACL injuries occur without contact during an activity such as a jump landing, biomechanical analysis can help determine whether or not someone has poor mechanics that are amenable to being corrected.

The most common mechanisms of ACL injury are hyperextension and valgus stress. Ireland emphasized the importance of recognizing athletes who land in the “position of no return” and teaching them to land in the “safety position” (see Figure 1). The “position of no return” involves trunk rotation to the opposite side, hip adduction and internal rotation, knee valgus and decreased knee flexion, internal tibial rotation, and foot external rotation. Instead, one should land from a jump with the trunk in a neutral position, the knee flexed without tibial rotation, and the center of gravity balanced over both feet. Because external ankle support affects the ROM and joint stiffness at the ankle, it is possible that someone could exhibit certain characteristics of landing in the “position of no return” when wearing external ankle support.

**Kinetics.** The center of mass (COM) is where an individual’s resultant GRFs are directed during activity. This is typically found within the trunk and will shift based on the placement of the body and alignment of the extremities while completing a given activity. As an individual’s control over trunk motion improves, the GRFs are better dissipated throughout the lower kinetic chain during movement. GRFs follow the COM along the X, Y, and Z axes. Lateral trunk flexion moves the COM so that the GRFs are dissipated just lateral to the knee joint, causing a valgus stress and risking the integrity of the ACL. A similar result occurs if the trunk does not move into enough flexion because
when coupled with decreased knee flexion, the GRF travels posterior to the knee and creates an external force that moves the tibia anteriorly, shearing the ACL.\textsuperscript{44,45}

\textbf{Figure 1.} The “position of no return” compared to the “safety position.” Adapted with permission from Ireland ML. Anterior cruciate ligament injury in female athletes: Epidemiology. \textit{J Ath Train.} 1999;34(2):150-154. Copyright 1999 National Athletic Trainers’ Association, Inc.

The hamstrings are responsible for preventing unwanted anterior tibial translation; however, the quadriceps counteract the external knee flexion moment.\textsuperscript{46} Landing in a more erect posture has shown greater electromyographic (EMG) activity of the quadriceps and a larger GRF than landing with trunk flexion.\textsuperscript{46} This results in decreased trunk and knee flexion, thereby causing more anterior tibial shearing force by increasing the patellar tendon-tibial shaft angle.\textsuperscript{9} Blackburn and Padua\textsuperscript{46} found that increasing trunk flexion during a jump landing reduces the vertical GRF and quadriceps EMG activity.
This is most likely due to the fact that trunk flexion moves the COM closer to the knee and decreases the moment arm (see Figure 2). With a smaller moment arm, the quadriceps is at a mechanical disadvantage, decreasing its effectiveness of moving the tibia anteriorly under the femur, which, in turn, helps protect the ACL.

Figure 2. Effect of trunk flexion angle on COM location. (A) Demonstrates the decreased quadriceps moment arm with greater trunk and knee flexion whereas (B) shows an increased moment arm when landing in an erect posture. Adapted with permission from Blackburn JT, Padua DA. Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity. *J Athl Train.* 2009;44(2):174-179. Copyright 2009 National Athletic Trainers’ Association, Inc.

One study found lateral trunk flexion to be a better predictor of ACL injury in females with 100% sensitivity and 72% specificity compared to trunk flexion. This is most likely due to knee valgus occurring within the same plane. It is possible that in the present study, having a single ankle braced or tape will disturb one’s proprioception. The
ankle support could act as a perturbation and affect lateral trunk displacement, thus increasing the number of errors committed on the LESS.

**Trunk and hip kinematics.** The interaction between the trunk and knee can be better understood when considering the lumbopelvic hip complex that connects them. Hip flexion coupled with knee flexion during a jump landing prevents anterior shearing forces on the ACL. It is when the hip does not flex fast enough following the joint compressive force attenuation that the ACL will most likely fail.\(^45\)

Frank et al.\(^47\) investigated the relationship between triplanar trunk and knee motion throughout the lumbopelvic hip complex during a side-step cutting task. They found that ACL loading mechanisms within the frontal and transverse planes are associated with trunk and hip biomechanics at the lumbopelvic hip complex. Trunk rotation combined with hip adduction is associated with frontal plane knee loading due to an increased external knee valgus moment.\(^47\) Thus, to better protect the ACL, one should have more trunk rotation in the direction away from a cutting maneuver in order to decrease the moment arm toward the lateral knee. Hip adduction extends that moment arm rather than shortening it.\(^47\) Therefore, the lumbopelvic hip complex, which incorporates movement at both the trunk and hip, directly influences biomechanics at the knee within the frontal plane.

The main contributor to ACL injury in the frontal plane at the knee is hip adduction and internal rotation because these movements place the knee in a valgus position. The valgus stress can be present either statically as the postural variant of an increased Q-angle or dynamically from poor neuromuscular control and strength.
According to Sigward, Ota, and Powers,\textsuperscript{24} the greatest predictor of frontal plane knee motion is limited hip external rotation. As a result, individuals with limited hip external rotation are likely to perform activity in an internally rotated position since there is more internal rotation available. This limited ROM at the hip coupled with poor strength of the hip abductors leads to a valgus alignment during dynamic activities because little prevents the knee from moving into this position.\textsuperscript{14,45}

**Foot and ankle kinematics.** Although the trunk and hip are considered to be large contributors to the kinematics of the knee joint, force absorption begins with the foot. The foot is the first part of the body to come into contact with the ground during activity and is therefore responsible for how the GRFs are dissipated throughout the body.\textsuperscript{6,25} Similar to how the posterior chain muscles at the knee are responsible for dissipating the GRFs, the ankle plantar flexors play a role in absorbing these forces initially.\textsuperscript{6} More specifically, Fong, Blackburn, Norcross, McGrath, and Padua\textsuperscript{6} found that the gastrocnemius is more influential than the soleus at attenuating forces due to its role in knee flexion. If more sagittal plane displacement is available, the loading phase will last longer and allow the forces to be dissipated over a longer period of time.\textsuperscript{6} As described earlier, knee flexion causes the hip to flex due to joint compressive force attenuation; ankle DF is crucial in allowing this to happen.\textsuperscript{6,45} When DF is limited, the rest of the kinetic chain is affected and the ACL is at a potentially increased risk for injury.\textsuperscript{6,24,25,48–50}

In the sagittal plane, the amount of DF available influences knee and hip flexion.\textsuperscript{6,50} When landing heel-to-toe, which is considered to be an error on the LESS,
there is less ankle DF available to absorb the GRFs compared to when landing forefoot first. This lack of DF directly decreases the amount of hip and knee flexion that can occur.\textsuperscript{50} Decreased motion also alters the lever arms of the plantar flexors, thus limiting the propulsive force available for push-off during activity, especially when changing direction.\textsuperscript{3} This difficulty when changing direction can adversely affect the trunk’s motion. As previously discussed, there should be more trunk rotation toward the new direction of movement; with less propulsive force, this is less likely to happen.

Simultaneously, with less strength to push off during a cutting maneuver, the external knee valgus moment will also be greater.\textsuperscript{47} All of these findings, however, do not agree with those of a study\textsuperscript{25} where the ankle DF was restricted by using a prophylactic ankle brace. The limited motion at the ankle was compensated through increased knee flexion. It is possible that those with limited DF incorporate other methods of absorbing the GRFs and downwardly displace the COM in order to protect the integrity of the ACL.

Ankle dorsiflexion also influences knee motion in the frontal plane because it is considered to be one of the main contributors to predicting knee valgus.\textsuperscript{24} Decreased DF prohibits the tibia from moving forward during a jump landing and single leg squat.\textsuperscript{24,42} In order to decelerate the COM from displacing downwardly, compensation may occur through either internal hip rotation or foot pronation.\textsuperscript{24,42} These two motions influence the knee by creating a valgus displacement and stressing the ACL. Hagins, Pappas, Kremenic, Orishimo, and Rundle\textsuperscript{48} found this relationship to be true when having participants land on either a flat or inclined surface. When on an incline that limited the amount of DF available, there was more knee valgus, as well as posterior GRFs, when
compared to the flat surface. Bell, Padua, and Clark\textsuperscript{49} reported a related finding when using a wedge under the calcaneus during a squat; increased DF limited the amount of valgus occurring at the knee. These studies portray how influential ankle motion is at the more proximal joints of the kinetic chain within both the sagittal and frontal planes and how alterations in ankle motion can put the ACL at a greater risk for injury.

Aside from the talocrural joint motion that influences knee movements, the subtalar joint motions of pronation and supination also play a key role in lower extremity biomechanics. Pronation is a combination of calcaneal dorsiflexion, eversion, and abduction while the talus performs the opposite motions in a fixed position.\textsuperscript{51} When weight-bearing, the talus affects the motions of the rest of the closed kinetic chain by forcing the tibia into internal rotation.\textsuperscript{38,51} The purpose of pronation is to absorb the GRFs during gait or jump landing; however, when this motion occurs in excess, the force is instead transmitted to the knee and tibia. As a result, the tibia is in an internally rotated position for a prolonged period of time, a situation that stresses the ACL. Beckett, Massie, Bowers, and Stoll\textsuperscript{38} reported that subjects with a history of ACL injury showed a greater navicular drop, which is a static measurement of pronation, in the affected limb when compared to an uninjured group. This finding suggests that excessive pronation may cause a preloading effect on the ACL, putting it at risk for non-contact injury.\textsuperscript{38}

Biomechanics in the lower kinetic chain play an integrative role for sports performance; therefore, it is crucial that athletes who present with abnormal motion at any or all of the joints be properly identified. Any of these motions can potentially be exacerbated if an external ankle support is applied, so the decision to use ankle taping or
bracing to prevent lateral ankle sprains may need to be considered on an individual basis.

Prevention of knee injuries begins with proper identification of those that are at risk.

Table 1 summarizes the risk factors associated with ACL pathology.

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Abbreviations: ACL, anterior cruciate ligament; CSA, cross-sectional area; DF, dorsiflexion

The Effect of External Ankle Support on Biomechanics

Inversion ankle sprains have a high incidence rate among athletes; therefore, it is a main focus of health care providers for prevention strategies.\(^1\,^2\,^4\) The combination of ankle inversion and plantar flexion injures the anterior talofibular ligament; sometimes the calcaneofibular ligament is also involved. Intervention strategies aim to prevent this mechanism of injury by reducing excessive inversion ROM with the use of external ankle support. Therapeutic exercises are used in conjunction to strengthen the muscles that dorsiflex and evert the ankle.

There are many types of external ankle supports used in the clinic to prevent inversion ankle sprains; these include a semirigid brace, a lace-up brace, the basket weave taping technique, and tape with moleskin reinforcement. Semirigid ankle braces are the most effective among these at restricting inversion and eversion, both before and after a period of exercise, by better than 45\%.\(^4\) The semirigid brace is not designed to restrict plantar flexion and DF; yet, the lace-up brace and tape applications do have an influence on sagittal plane motion.

The lace-up brace influences plantar flexion ROM due to its material support over the anterior ankle and midfoot.\(^4\) Meanwhile, the basket weave technique and moleskin reinforcement require the foot to be placed in DF during application of the stirrups, which inherently decreases the amount of plantar flexion available. Reinforcement from the moleskin restricts ankle ROM significantly more than the basket weave application by itself.\(^26\) Although all of these interventions affect the amount of plantar flexion ROM
available following application, understanding DF restriction is more pertinent as to how it affects lower extremity biomechanics when landing from a jump.

Dorsiflexion ROM is restricted more with tape application compared to the lace-up brace.\textsuperscript{4,5} Normally tape loses its mechanical and adhesive restrictive properties after 20 minutes of exercise. The tape fibers start to separate or tear and perspiration affects the adhesive property, all decreasing its ability to restrict ROM.\textsuperscript{52} With Purcell, Schuckman, Docherty, Schrader, and Poppy\textsuperscript{5} finding that tape can still restrict ankle ROM by as much as 65% following exercise, the adverse effects of poor force absorption are prolonged during dynamic activity such as jump landing.

The literature is conflicted regarding the effect that external ankle support has on the knee. DiStefano, Padua, Brown, and Guskiewicz\textsuperscript{25} expected that during a jump landing, knee flexion would decrease and vertical GRF would increase when DF was restricted while wearing a lace-up ankle brace. Instead there was an increase in knee flexion and vertical GRF was not affected. This is most likely due to a compensation for the restrictions placed at the ankle in order to keep the vertical GRF constant between conditions.\textsuperscript{25} This study shows that wearing an ankle brace would reduce the risk of ACL injury by increasing the degree of knee flexion during a jump landing.

As for the effect that a lace-up brace places on the knee within the frontal plane, studies\textsuperscript{27,34} have found no influence on varus or valgus torque. Venesky, Docherty, Dapena, and Schrader\textsuperscript{34} found an effect on transverse plane biomechanics at the knee while wearing a lace-up brace. External rotation torque was greater when wearing the brace during a drop-landing on a slanted surface that simulated an inversion mechanism.
More force would then be needed to prevent internal rotation from occurring at the knee, which is one of the components of an ACL injury.

A brace or taping application limits excessive motion not only at the talocrural joint but also at the subtalar joint where rotation occurs. Because the goal of external ankle support is to prevent inversion, it places the subtalar joint in a position that includes plantar flexion and internal rotation of the talus. The tibia is fixed upon the talus and thus performs the same motions; therefore, it is inherently forced into internal rotation and places the ACL at risk for injury.\(^{34}\) The findings by Stoffel et al.\(^ {27}\) do not agree with this because they found the peak internal rotation moment at the knee to actually be reduced during a sidestepping maneuver. The lack of consistency among these studies provides further evidence that more research is needed regarding the effect that alterations to ROM at the ankle have on the knee during activity.

One result that has been consistent among studies is that the lace-up ankle brace and the basket weave taping technique reduce ROM at the ankle. Since DF is the most important at attenuating forces during a jump landing, its reduction may influence biomechanics at the knee. With inconsistent results demonstrated thus far, it may be more beneficial to directly determine whether or not an ankle brace or tape application increases or decreases the risk of an ACL injury through the implementation of the LESS. Comparing passive ankle DF stiffness between the two interventions will presumably show that there is actually reduction in ankle joint motion that can be attributed to the tape or brace. Then, instead of having to interpret biomechanical results of a jump landing to determine the effect that these interventions place on the ACL, measuring a
difference in this risk-assessment tool will provide distinguishable results. Understanding the effect that external ankle support has on the knee will provide useful information for athletes and medical professionals regarding the potential risk of injuring the ACL when they attempt to protect the ankle.

**Evaluating Lower Extremity Biomechanics**

**Ankle dorsiflexion range of motion.** The amount of dorsiflexion available during activity depends on both arthrokinematic and soft tissue restrictions. The Achilles tendon is one of the integral factors in jump-landing kinetics because of its elasticity. Measuring ankle DF with a goniometer can accurately determine the flexibility of the gastrocnemius and soleus; however, measuring passive joint stiffness is more reflective of soft tissue extensibility when weight bearing or landing from a jump. Kawakami, Kanehisa, and Fukunaga found a positive correlation between Achilles tendon stiffness and passive plantar flexion torque, indicating that stiffness is one of the limiting factors to joint flexibility. In fact, they found that 70% of the variance of passive torque can be explained by the variance of tendon stiffness among individuals. It is unknown how much of the joint stiffness can be attributed to joint compliance when there are other surrounding soft tissue structures involved, but it can be assumed that the triceps surae is the main contributing factor.

A more compliant Achilles tendon transfers less energy to the proximal structures; therefore, if there is decreased compliance in the joint during DF, more energy is transferred to the knee during a jump landing. Whitting, Steele, McGhee, and Munro investigated the relationship between passive DF stiffness and kinetic data but
were unable to find any significant results regarding vertical GRF, peak DF angle, and peak DF moment during a single-leg drop landing. The results demonstrated that a higher Achilles stiffness does not negatively alter landing biomechanics. The authors attributed these insignificant findings to the difficulty and variability of landing on a single foot instead of a difference in Achilles stiffness. More research is needed to determine the relationship between passive ankle DF stiffness and landing biomechanics, but joint stiffness should be evaluated as a factor in flexibility and torque.

**Measuring passive ankle dorsiflexion stiffness.** Torque increases as a joint is moved passively, so using an isokinetic dynamometer to measure how much a joint moves with an applied force is considered the gold standard. Unfortunately, most machines are costly, bulky, and time-consuming for users, making it impractical for easy use in the clinical setting. Another method of measuring ankle dorsiflexion stiffness using the Lidcombe template was proposed in 1991 by Moseley and Adams. It was designed so that a known force could be applied to the joint and the degree of movement could be easily determined, without needing an expensive isokinematic dynamometer.

The template involves a spring scale and a Plexiglass plate ruled with parallel lines. The person is positioned with its foot within the template and the knee slightly flexed. The spring is wrapped around the forefoot at the heads of the metatarsals and is stretched to apply varying amounts of force to the talocrural joint. The parallel lined Plexiglass is used to determine the degree of DF experienced with each increase in load. This angle is measured using a protractor on a photograph of the foot in the Lidcombe template once maximum passive DF is achieved. The results of the Lidcombe template
has excellent interrater reliability (ICC = 0.97) but there are no data verifying the validity of this measurement tool.\textsuperscript{29}

Since the introduction of the Lidcombe Template, a modified version of this tool has been developed to address its limitations. Because the original version placed the knee in a flexed position, there was limited influence by the compliance of the gastrocnemius muscle during passive DF.\textsuperscript{30} Thus, the procedure did not reflect the true stiffness of the triceps surae muscles and Achilles tendon on this motion. In addition, without a stiff board that fixes the position of the foot, the motions occurring in the frontal and transverse planes such as inversion/eversion and pronation/supination, can alter the absolute influence of the triceps surae and Achilles tendon on just dorsiflexion.\textsuperscript{30}

The modified Lidcombe template (see Figure 3) uses two boards that are connected by a hinge to serve as the axis of rotation. The strain gauge is attached 200 mm from the hinge on the foot plate, setting a fixed position for the lever arm when torque is applied to the joint. An inclinometer is attached on the underside of the foot plate for a more accurate measurement of dorsiflexion compared to the indirect measurement of using a protractor on a photograph.\textsuperscript{30}

The modified Lidcombe template has excellent interrater reliability (ICC = 0.995, SE = 0.63°).\textsuperscript{30} Again there are no data regarding the validity of this measurement tool; however, it is very difficult to obtain a highly valid measurement of ankle dorsiflexion with any tool because of the complexity of the talocrural joint. Goniometry measuring ankle DF with the knee extended and flexed also has a high reliability (ICC = 0.821, 0.904) but a larger standard error (SE = 3.60°, 2.50°).\textsuperscript{42} Therefore, face validity is the best
validity that can be acquired for the modified Lidcombe template.\textsuperscript{30} Based on the ease of execution and reliable results produced by the modified Lidcombe template, this is the best option for measuring passive ankle DF stiffness.

Figure 3. The modified Lidcombe template. The foot is secured within the template while the participant pulls on the spring scale, moving the foot into dorsiflexion. A digital inclinometer is placed against the outer edge of the foot board to measure the angle of displacement. Photographs by the author.

Three-dimensional motion analysis. Three-dimensional motion analysis in a laboratory setting is considered the gold standard for evaluating the kinematics and kinetics of lower extremity movements.\textsuperscript{18,21,55} It is most commonly used in gait analysis, but its ability to detect small changes make it useful for analyzing virtually any dynamic activity.\textsuperscript{56} This is helpful when determining whether or not an intervention such as external ankle support has any effect on lower extremity biomechanics.

Electromagnetic tracking devices such as Ascension Technology’s “Flock of Birds” (Burlington, VT) are used often for biomechanical analyses.\textsuperscript{57} These measure the
position and orientation of various sensors placed throughout the kinetic chain with respect to a single transmitter. Using direct current magnetic fields, the transmitter then orients the sensors within space and digitizes the joints. The manufacturer suggests that this tracking device is accurate at detecting changes in position of as little as 2.5 mm and 0.5° of rotation. Yet, Milne, Chess, Johnson, and King found the device to have sensitivity high enough to detect changes as small as 0.25 mm and 0.1°. Its positional and rotational errors were within 2% when operating within the device’s optimal range of 22.5-64.0 cm from the transmitter.

Another tool that has been used for recording 3-D kinematic data is the Vicon motion analysis system (Oxford Metrics Ltd., Oxford, UK). It captures optoelectronic motion of retroreflective markers that are recorded by several cameras. The cameras have LED strobe lights surrounding their lenses and capture the reflected light from the markers, recording the markers’ positions. The accuracy of the Vicon system is very good with errors of only 63 μm. With results this precise, it is evident why 3-D motion analysis is considered to be the gold standard for collecting kinematic data.

Although valid data are imperative when choosing a method for evaluating biomechanics, reliability is equally important. Reliable data for gait analysis are dependent on the ability to detect small changes across different sessions following an intervention. Based on a review of many studies, McGinley, Baker, Wolfe, and Morris found that the most reliable data are obtained from measurements of motions within the sagittal plane (ICC = 0.80, SE = 4°). Measurements in the coronal plane had an ICC greater than 0.70 and a standard error around 2°, while the ICC in the transverse plane
was less than 0.70. Within the transverse plane, the measurement of hip rotation yielded
the worst reliability; however, since the standard error was still less than 5° for most
studies, it may still be considered reliable since relatively low error is still achievable.
McGinley et al. found that most studies also reported good interrater reliability between
sessions.

Reliability is affected by errors that can be either intrinsic or extrinsic. Intrinsic
errors, such as how an individual’s gait changes inherently, produce variation that is
inevitable among trials. Extrinsic errors are procedural and should be closely monitored
in order to maintain valid and reliable results. These errors include inconsistent
placement of the markers over specific landmarks, variation in walking speed between
trials, data processing, and equipment measurement errors. The markers are susceptible
to movement during trials since they are placed on the skin instead of being fixed on the
landmark. Although 3-D motion analysis may involve some error, it is still considered the
gold standard because it produces the most accurate results for evaluating biomechanics.

While using 3-D motion analysis would be ideal when determining whether or not
a patient displays the biomechanical risk factors associated with ACL pathology, it is not
practical for the clinical setting. Screening for ACL injuries during the preseason needs to
be fast, easy, and inexpensive while still producing valid results. Two-dimensional (2-D)
motion analysis solves the financial, special, and temporal costs associated with 3-D
systems. A 2-D approach requires use of standard video cameras that are not as accurate
as 3-D, yet it is better suited for clinical use. If it provides results that are valid and
reliable enough when compared to 3-D systems, it would be better for screening patients and the associated risk of an ACL injury.

A study by McLean et al.\textsuperscript{55} aimed to determine whether or not the measurement of 2-D motion within the frontal plane was consistent with the results of a 3-D system. The data were collected simultaneously during a series of activities including a side jump, side step, and shuttle run. The frontal plane knee angles were estimated from the video tape for manual digitization of the joint center. Capturing data in 2-D was time consuming and not nearly as accurate because, overall, the angles recorded using the video cameras were consistently greater than the 3-D measurements. Nonetheless, it can still be used as a screening tool because the results were reliable and provided moderate concurrent validity since the mean peak angles correlated well with those of the 3-D system.\textsuperscript{55}

A 2-D system would be better suited for recognizing larger differences in knee valgus rather than small changes that require more precise measurements. While McLean et al.\textsuperscript{55} were able to determine that using video cameras is a more efficient screening tool for ACL injury risk when compared to 3-D motion analysis, it only provides reliable results for knee valgus. In order for 2-D analysis to be considered a better choice for a screening tool than 3-D, it must be able to validly and reliably analyze all of the associated biomechanical risk factors for ACL injury.

The Landing Error Scoring System. The Landing Error Scoring System (LESS) was developed as a practical assessment tool for identifying high-risk movement patterns during a jump landing.\textsuperscript{18} It uses two video cameras to capture multiple trials of the task in
order to determine how many errors someone commits during a jump landing. The errors are based on the biomechanical risk factors associated with ACL injury within the sagittal, frontal, and transverse planes throughout the trunk and lower extremity. The LESS is a good option for use in the clinical setting because it is inexpensive, provides fast results, is easy to use, and provides both valid and reliable results.\textsuperscript{18}

To complete the jump-landing task for the LESS, participants must jump a distance that is 50\% of their height away from a 30 cm high box and immediately jump back up to a maximum height after landing.\textsuperscript{18} In order to be considered a successful trial, they must jump and land with both feet in a fluid motion. The jump off of the box must also be a forward jump towards the landing area instead of a vertical hop downward. Three trials are completed for scoring and the procedure takes a mere 5 minutes per participant to complete.

The LESS scores are determined based on the number of errors exhibited during the jump landing. Two video cameras are used to record the trials; one camera captures movements within the frontal plane while another one is placed to capture the sagittal plane. Each camera is set up 3.45 m (136 in) from the force plate and 1.22 m (48 in) off the ground.\textsuperscript{18} The rater needs only needs 5 minutes to watch the videos from each camera and score the three trials for each person. Hence, only 10 minutes would be needed for someone to complete the task and then provide a score, making it faster than setting up sensors for 3-D motion and analyzing the results.

The score for the LESS depends on 17 items that assess the position of the lower extremity and trunk at initial contact with the ground as well as movements at the point of
maximum knee flexion. There also are two global items that consider overall movement and impression of the jump landing quality. Scores for each item can range from 0-2; since the test is assessing the presence of errors, a higher score indicates an increased risk for ACL injury. The scoring sheet with each item and the possible scores is provided in Appendix A.

In order to determine the validity of the LESS as a screening tool for ACL pathology, the overall scores were compared with kinetic and kinematic measurements taken by a 3-D motion analysis system. Padua et al. established concurrent validity by dividing participants into four quartiles to represent the quality of their jump landings. The overall scores were also discriminately valid based on sex since females have a higher risk of ACL injury compared to males. The LESS’ ability to significantly distinguish the scores across the different groups and between sexes suggests it is a valid tool for assessing ACL injury risk.

Statistical analyses of the individual LESS scores compared with joint kinematic and kinetic data revealed nearly every variable to be correlated with the LESS scores including: decreased knee and hip flexion angles, increased knee valgus and hip adduction angles, increased internal rotation moment at the hip and knee, increased hip and knee extension moments, increased knee and hip adduction moments, and increased peak vertical GRF. Some of these were different among all quartiles, while others were different only between the poor and excellent groups. The importance of these findings is that the LESS can be accepted as a tool to assess multiplanar biomechanical risk factors.
of ACL injury for the clinical setting, especially for differentiating between males and females and between those that score high or low.\textsuperscript{18,21}

Good reliability of the LESS is also important for its implementation in the clinic. Raters undergo a comprehensive training program full of descriptors on the importance of proper biomechanics and examples of how certain errors can be identified with the LESS. They must score 20 subjects and compare the results to an expert rater with any discrepancies being clarified in order to achieve as much consistency as possible.\textsuperscript{18} Interrater reliability for the LESS was good with an ICC and SE of 0.84 and 0.71, respectively.\textsuperscript{18}

In order for a system such as the LESS to be feasible throughout most clinics, the scores need to be reliable across different levels of expertise within the sports medicine environment. Onate, Cortes, Welch, and van Lunen\textsuperscript{21} established excellent novice versus expert interrater reliability. All raters were certified athletic trainers. Experts had at least 15 years of experience, while novices had less than 1 year. The experts were also involved in the development of the LESS scoring evaluation form, making them more experienced not only in the clinical setting but with biomechanical analysis of ACL injury risk. There was excellent reliability for the overall scores of the LESS between the two groups (ICC = 0.835).\textsuperscript{21} This level of reliability can be most likely attributed to the fact that video cameras are easy to use and the movement descriptors are dichotomous instead of requiring a visual estimation of joint angles like those used in previous 2-D systems.\textsuperscript{21}
Some of the individual items, as scored by the raters in the study by Onate et al.,\textsuperscript{21} did not have significant agreement with the results of the 3-D motion analysis. One item that stood out with poor agreement of 21\% between raters and the 3-D system was the knee flexion angle at initial contact.\textsuperscript{21} Since the amount of knee flexion is considered to be one of the more important biomechanical risk factors based on the shearing force it places on the ACL,\textsuperscript{44,45} the LESS shows room for improving the validation of some items on the scoring sheet.\textsuperscript{21} Neither has the LESS also shown good predictive validity for ACL injuries. For example, one study\textsuperscript{33} analyzed more than 5,000 college and high school athletes over the course of 3 years and only 32 participants with a high score on the LESS suffered an ACL injury. These results are limited as more ACL-injured athletes must be screened with the LESS in order to have sufficient statistical power. The LESS still holds the potential to be effective in a clinical setting, but more studies are needed to further confirm its validity.

The LESS has been modified specifically for the clinical setting by turning it into a tool that can be used for real time analysis without the use of video cameras. The Landing Error Scoring System-Real Time (LESS-RT) has better utility for clinicians because it is even faster and easier for preseason screenings than the original LESS. The execution of the LESS-RT is the same except that it requires four trials for evaluation and only 10 items are scored.\textsuperscript{9} Two trials are completed with the rater directly in front of the participant, analyzing frontal plane motion, while two more trials are conducted with the rater standing to the side and analyzing sagittal plane motion. It has similar reliability (ICC = 0.72–0.82, SE = 0.69–0.79) and validity of finding differences between those
with high and low scores. Further research is required before the LESS-RT can be fully implemented in the clinical setting.
Chapter 3: Methods

Participants

A convenience sample of 28 participants were recruited to participate in this study. The sample was taken from the intercollegiate and club sports teams at Ohio University. Participants were recruited with an e-mail from their athletic trainer or club sport president. Once enrolled in the study, they read and signed an informed consent form prior to data collection. As a within-subjects repeated measures study, each participant acted as its own control and received both the lace-up brace and taping conditions while completing the jump-landing task

Inclusion and exclusion criteria. To be considered eligible for this study, participants had to meet the following criteria:

1. Between the ages of 18 and 28 years.
2. Member of one of the intercollegiate or club sports teams at Ohio University.
3. Not experiencing symptoms from a lower extremity or back injury within the three months prior to participation.
4. No previous history of ACL reconstruction or lower extremity surgery.
5. Not using external ankle support for the majority of practices and games.

Power analysis. The pilot study on the LESS by Padua et al.\textsuperscript{18} divided the overall scores into quartiles defined as excellent, good, moderate, and poor. Based on descriptive statistics, it was assumed that the majority of subjects would fall into the middle two quartiles of good and moderate. With the hypothesis that LESS scores will increase with application of external ankle support, the scores were expected to shift toward the
quartiles of moderate and poor. Using the mean scores and standard deviation of each group, the effect size was calculated to be 0.21. A power analysis was conducted with an effect size of $n^2 = 0.25$. Using a 1 group by 3 measures model where $r_B = 0.80$, $r = 0.50$, and $\alpha \leq 0.05$, the number of participants needed was 28.

**Procedures**

Participants reported to the Biomechanics Laboratory in Grover Center, Room E116 for the testing session. A within-subjects repeated measures study in a controlled laboratory setting was conducted to assess passive ankle dorsiflexion stiffness and jump-landing biomechanics using the LESS in all three of these conditions on each participant within the same testing session: control (i.e., no ankle support), lace-up ankle brace, and basket weave taping technique. Participants’ height and weight were recorded before beginning the testing session.

**Passive ankle dorsiflexion stiffness.** To determine whether or not there was an actual change in ankle DF when applying either the lace-up ankle brace or tape, ankle DF stiffness was calculated using the modified Lidcombe template (see Figure 3). The template consists of a baseboard that is positioned underneath the triceps surae and a foot plate that runs along the plantar aspect of the foot. The boards are connected with a hinge, creating the axis of rotation during DF. The foot plate is 32 cm in length to accommodate up to a US size 15 foot. A spring scale (Model no. 8910, Hanson, Shubuta, MS) was connected to the base of the foot plate, 20 cm from the hinge, in order to measure the application of amounts of force on the ankle joint. A digital inclinometer (Acumar Digital Inclinometer, Model ACU001, Lafayette Instrument, Inc., Lafayette, IN) was attached to
the bottom of the foot plate to measure the angular changes of the joint as force is increased.

Participants long sat on a table with their trunk against the wall and the knee extended while the foot was affixed in the template. No measurements were taken in a knee-flexed position. The testing began with the ankle at a position of 90°, which is considered neutral. The amount of force applied was increased at increments of 5 lb and the angle that the footplate moved from neutral, as measured by the inclinometer, was recorded at each increment.

**The Landing Error Scoring System.** Two standard video cameras (Logitech, Newark, CA) captured each trial to be scored by the LESS. The cameras captured footage from both the frontal and sagittal plane during the trials. Each camera was placed 3.45 m from the landing area and 1.22 m off the ground. The landing area was a 1.22 m square designated by white athletic tape and remained in the same location for each subject. The 30 cm box where the task initiated was moved away from the landing area a distance that was 50% of the participant’s height.

The jump-landing task consisted of two segments: landing from a 30 cm box and a vertical jump. Upon their initial landing (see Figure 4C), the participants were instructed to immediately rebound to a maximal vertical jump height (see Figure 4E). Each participant was allotted three practice trials to become familiar with the task. A successful task is characterized by jumping off of the box with both feet, jumping forward but not upward during the initial segment (see Figure 4B), and completing the task in a fluid motion. Participants did not receive feedback or prompting following the
initial instruction of the task unless the procedure was incorrect. Three successful testing trials were recorded for data collection for each of the three conditions per participant, for a total of nine trials. Application order of the independent variables was randomized using a table of random numbers.

![Figure 4](image)

**Figure 4.** Progression of the jump-landing task. (A) Jumping off a 30 cm box (B) Horizontal jump (C) Initial contact (D) Maximum displacement (E) Maximum vertical jump. Photographs by the author.

**External ankle support.** The ankle of the dominant foot of each subject was fitted for either a small, medium, large, or extra-large ASO brace (Medical Specialties, Charlotte, NC) as recommended by the manufacturer on the dominant foot. The dominant limb is defined as the limb preferred when kicking a ball.

Prior to tape application, the dominant foot was cleaned, then applied with tape adherent and prewrap. The participant sat on the table with the limb being tested extending past the edge of the table and the foot maintained at a 90° angle to be taped.
Pre-wrap extended from the musculotendinous junction of the triceps surae to the midfoot. The basket weave taping technique described by Prentice⁵⁹ was used.

Anchor strips were applied below the belly of the gastrocnemius and over the styloid process of the fifth metatarsal around the midfoot. Beginning on the medial side, 3 stirrups were applied that began at the proximal anchor, pulled the ankle into eversion around the heel, and ended on the lateral side of the anchor. Each stirrup was alternated with a horseshoe that started at the distal anchor and went around the posterior heel to the other side, overlapping at least half of the preceding horseshoe. Next, 3 heel locks were applied to complete the technique. The heel lock began above the medial longitudinal arch and hooked around the lateral heel to finish at the starting point. The maneuver was reversed on the lateral side to complete the entire heel lock. Finally, the tape was closed with horseshoe strips applied up the ankle. Three testing trials of the LESS were completed following application. Once the task was completed, the tape was removed with a pair of scissors.

**Data Collection**

Demographic data including age, sex, height, weight, and sport were recorded on a data collection sheet. Measurements for DF stiffness and the LESS were taken on the dominant limb.

Passive ankle joint stiffness was calculated using the measurements obtained from the modified Lidcombe template. At increments of 5 lbs (2.27 kg) of force on the spring scale, the amount of DF achieved was recorded by the digital inclinometer; the starting point was ankle neutral. The resulting series of points were used to calculate an
exponential regression line and stiffness coefficient (see Figure 5). The stiffness coefficient was the amount of torque per degree of movement. Torque is defined as the amount of force as applied by the spring, multiplied by the moment arm. To account for each participant’s height, the moment arm was calculated based on the ratio of height to foot length. The amount of force applied was calculated as a percentage of the individual’s total body weight.

The total score of the LESS is calculated from 17 items listed in the scoring sheet of Appendix A. A trained evaluator watched each trial in order to calculate an overall score based on a standardized set of scoring criteria. The evaluator was able to use video pause and rewind controls in order to identify the proper frame for scoring. Scores are based on the dominant leg only. Items that are scored from the sagittal plane view can be scored together; the same can be done with the items from the frontal plane view.

Most items on the LESS were scored dichotomous yes (1) or no (0) based on whether or not each biomechanical error was present during execution of the task. Errors that would receive a score of 1 include: knee flexion at initial contact less than 30° and less than 45° of displacement, knee valgus, limited hip flexion, trunk extension, lateral trunk flexion, toes internally or externally rotated more than 30°, narrow or wide stance width, asymmetrical foot contact, and not landing toe-to-heel.
Joint displacement within the sagittal plane was scored 0, 1, or 2 if it was considered soft, average, or stiff, respectively. The overall impression of the jump landing, as considered by the trained evaluator, was scored on a similar scale so that excellent was 0, average was 1, and poor was 2. Since the LESS determines the number of errors occurring during a jump-landing task, a higher score indicates poor technique and a higher risk of injuring the ACL.

Statistical Analysis

All data were computed using SPSS statistical software (IBM Corp. Released 2013. IBM SPSS Statistics for Windows, Version 22.0. Armonk, NY: IBM Corp). Both dependent variables, the stiffness coefficient and overall LESS score, were compared via
a one-way repeated measures ANOVA ($\alpha \leq 0.05$) within subjects. The purpose was to
determine if joint compliance was influenced by application of the tape or brace using the
stiffness coefficient. The analysis of the overall LESS scores provided insight toward the
influence of the three conditions on jump-landing biomechanics for each individual.
Mauchly’s test of sphericity was computed to determine if the variances between all
group combinations were equal. Any significant main effects were evaluated post hoc
with a Bonferroni correction.
Chapter 4: Results

All 28 participants, 14 males and 14 females, completed the study. Repeated measures ANOVA with a Huynh-Feldt correction for within-subject effects ($F_{1.725, 46.575} = 0.328, P = 0.690$) revealed no differences in passive ankle dorsiflexion stiffness coefficients among the three conditions of no ankle support, ankle taping, and ankle brace (see Figure 6). The mean stiffness coefficient for each condition were $0.227 \pm 0.0665$, $0.226 \pm 0.0708$, and $0.238 \pm 0.122$, respectively.

![Stiffness Coefficient for each type of external ankle support.](image)

The LESS scores ranged from 0 to 12 for the control condition, 1 to 10 for the ankle taping, and 3 to 11 for the ankle bracing. Mean LESS scores for each of the conditions were $5.29 \pm 2.33$, $5.60 \pm 1.88$, and $6.10 \pm 1.96$, respectively (see Figure 7).
Repeated measures ANOVA with a Huynh-Feldt correction for both within-subject effects ($F_{1.658, 44.757} = 5.686, P = 0.009$) identified a significant main effect for how external ankle support affected overall LESS scores. Post hoc analyses with a Bonferroni correction ($\alpha \leq 0.016$) indicated that wearing an ankle brace resulted in more errors on the LESS when compared to the control ($P = 0.003$). There were no differences in LESS scores between the brace and the tape ($P = 0.048$) nor between the tape and control ($P = 0.959$).

**Figure 7.** Overall LESS scores by condition (* denotes significantly different from the control, $P = 0.003$).
Chapter 5: Discussion

The purpose of this study was to determine whether or not a decrease in ankle DF caused by wearing an ankle brace or tape increases the risk of ACL injury as determined by the Landing Error Scoring System (LESS). The results suggest that passive ankle dorsiflexion stiffness is not affected when wearing either an ankle brace or tape. Although passive stiffness was anticipated to provide insight about whether or not external ankle support affects an individual’s jump-landing biomechanics, the overall LESS scores did not support this because only the scores collected with participants wearing the brace increased significantly. Nonetheless, this important finding suggests that wearing an ankle brace can potentially lead to increased risk of ACL injury due to the presence of more biomechanical risk factors during a jump landing.

The most common errors committed while wearing the brace compared to the control were asymmetrical foot contact upon landing, lateral trunk flexion at initial contact, and foot external rotation greater than 30°. The asymmetrical foot contact was common particularly on the first of the three trials while wearing the brace because the participants likely were not accustomed to the brace’s restrictions. Participants landed either with the feet contacting the floor at different times or with one foot landing heel-to-toe while the other foot landed toe-to-heel. This finding is important primarily in individuals adjusting to wearing an ankle brace for the first time; it is unknown how prolonged use of an ankle brace would affect overall LESS scores.

Lateral trunk flexion toward the side on which participants wore the ankle brace occurred in several participants and did not differ among the first, second, or third trials.
This finding is relevant because lateral trunk motion changes the location at which the body’s COM is located during a task such as a jump landing and is considered to be a strong predictor of ACL injury.\textsuperscript{43,44} Lateral trunk flexion occurs in response to poor core stability; when an external perturbation such as an ankle brace is applied, the body must react appropriately in order to maintain a stable position. Based on this study, one should consider implementing a core stabilization program if choosing to wear a prophylactic ankle brace on a single limb in order to reduce the likelihood of ACL injury.

The most common jump-landing characteristic of participants in this study while wearing the brace was excessive foot external rotation. Not only is external rotation an error on the LESS, it is one of the motions included in Ireland’s “position of no return.”\textsuperscript{14} Many participants’ jump-landing biomechanics changed between the control (A) and brace (B) conditions (see Figure 8). The amount of foot external rotation increased while knee flexion remained relatively constant. Precise measurements are unattainable without 3-D motion analysis to measure joint angles and relative lower extremity motion; however, the risk for ACL injury using the LESS is based on a participant achieving less than 45° of knee flexion displacement. This angle did not change substantially across trials in any of the participants who exhibited this trend.

Compensation for limited dorsiflexion range of motion can manifest in many forms throughout the lower kinetic chain. Typically decreased dorsiflexion is associated with increased knee flexion, ankle inversion, and foot pronation.\textsuperscript{61} In those who land from a jump with the foot in an externally rotated position, pronation and internal tibial rotation will occur concurrently in order to obtain maximum flexion. One study evaluated
knee kinetics during a single-leg jump landing onto a slant board with a braced ankle. The researchers observed external rotation torque to be greater when wearing the brace than without. This was most likely due to the brace decreasing supination and tibial external rotation. Therefore, the brace allowed more tibial internal rotation and pronation to occur and the knee musculature had to counteract those motions with greater torque. The ACL becomes taut during internal tibial rotation and is therefore at risk for injury if the motion is excessive and the antagonistic muscles are weak.

Figure 8. Foot external rotation compensation. (A) Control (B) Brace. Note the outward rotation of the feet in (B). Photographs by the author.

Foot pronation occurs in response to the lack of available ankle dorsiflexion in order to achieve adequate knee flexion during a jump landing. Of the participants who landed with more foot external rotation while wearing the brace compared to the control condition, the majority of them still did not exhibit the error of knee flexion displacement less than 45°. In order to achieve the same amount of knee flexion when wearing the
brace, participants may have compensated with pronation, which presented as foot external rotation. This finding is consistent with the findings of one study that reported an increase in knee flexion angles in response to decreased DF while wearing a brace.\textsuperscript{25} Knee flexion is considered one of the strongest predictors of ACL injury, but it is only one of many.\textsuperscript{43–45} A single risk factor may not increase the risk of ACL injury. Rather, the combination of multiple biomechanical factors increase the chance of injury.\textsuperscript{18} While pronation allows for better absorption of GRFs during landing instead transmitting the forces to the knee, this attempt to protect the knee still is not enough to prevent manifestation of other risk factors.

Aside from wearing the brace, we hypothesized that the ankle tape would also affect ACL injury risk. Tape has a negative effect on dynamic movement performance by decreasing vertical jump height and slowing performance based on its restrictions in all three planes of movement.\textsuperscript{26} Tape also decreased DF during a sidestepping maneuver, resulting in the reduction of both internal rotation and varus moments at the knee.\textsuperscript{27} These findings suggest that decreased DF does not increase knee joint loading and the risk of ACL injury. This supports findings that the GRF remained constant when wearing an ankle brace during a jump landing.\textsuperscript{25} While knee kinetics may be altered when the ankle is taped, the effect is not large enough to become apparent through the LESS.

Neither the tape nor the brace appeared to have an effect on medial knee displacement during the jump landing even though knee valgus is one of the most common mechanisms of non-contact ACL injury. Evidence shows that individuals with decreased DF experience increased knee valgus while squatting and landing from a
jump. However, a decrease in DF associated with wearing external ankle support does not appear to have the same effect. Evidence shows that there is no effect on knee varus or valgus torque during a single-leg drop-landing on a slant board as well as no increase in knee valgus while wearing a brace during a jump landing. It is possible that in order for limited DF to have an effect on knee valgus, the restrictions must be due to soft tissue instead of external ankle support. Internal restrictions have more control over skeletal movements than external restrictions. Further inquiry is needed to determine which has a greater effect on altered movement patterns at the knee in the frontal plane.

Based on measurements with the modified Lidcombe template, there were no significant differences in passive ankle dorsiflexion stiffness among conditions. Limitation of ankle motion by the brace still occurred during the jump-landing task even though the statistics did not indicate a difference in its stiffness. This change was not only reflected by the overall scores on the LESS, many participants visibly exhibited less DF when going through maximum flexion displacement during the landing. When not wearing any external ankle support, the heel maintained contact with the ground and there was greater ankle DF when landing (see Figure 9A). Meanwhile, while wearing the brace, the heel of the same participant did not contact the ground because the ankle was not able to achieve as much DF (see Figure 9B). While the results of passive ankle dorsiflexion stiffness among the conditions may not have been significant, they did suggest a tendency for the brace to increase stiffness more than the tape did.
Although our original hypothesis stated that passive ankle dorsiflexion stiffness would be affected by the tape and brace as measured with the modified Lidcombe template, this was not the case. However, participants still committed more errors on the LESS while wearing the brace, thus indicating that the ankle brace did have an effect on jump-landing biomechanics by restricting DF. Whether it is through compensation or poor adjustments to the external perturbation of the brace, wearing a prophylactic ankle brace should be considered as a contributor to one’s jump-landing alterations that could potentially jeopardize the ACL. The results of this study are consistent with the literature and provide insight to help players, parents, coaches, and clinicians make evidence-based decisions about whether wearing prophylactic ankle braces to protect the lateral ankle ligamentous structures is worth the increased risk of ACL injury.
Limitations

There are a number of limitations regarding the use of the modified Lidcombe template that may have led to insignificant results for passive ankle dorsiflexion stiffness. Neither the original Lidcombe template nor the modified version were subject to validation of their results and their face validity was presumed. Isolating motion at the talocrural joint is difficult to achieve since the subtalar joint also contributes to ankle dorsiflexion. Therefore, any measurement of ankle DF ROM and passive stiffness will produce some error.

Ankle dorsiflexion range of motion is limited by the stiffness of multiple posterior muscles and connective tissues. For this study, measurements were only taken with the knee in an extended position in order to account for how the gastrocnemius contributes to limited motion. However, measuring motion in this position does not translate to the completion of a dynamic task because such movements do not occur only with the knee extended. When landing from a jump, knee flexion is encouraged in order to better attenuate the GRFs.

Recording measurements with the knee in full extension, instead of including those while the knee was flexed, could have limited the results of the modified Lidcombe template. The original Lidcombe template took measurements only with the knee flexed, which eliminated any influence from the gastrocnemius and the modified version sought to avoid that limitation. Future studies measuring passive ankle dorsiflexion stiffness should consider measuring in both positions in order to account for a position that is more functional, as well as considering the role of the gastrocnemius.
It is likely that in order to achieve more DF, many participants compensated with foot pronation, ankle inversion, and/or knee flexion. Upon initiation of the measurements, each participant was instructed to maintain the knee in full extension and not to move the foot against the board in any way. Nonetheless, it appeared that foot pronation and ankle inversion were inherent in many participants’ performances and may have altered the measurements. This possibly could have been avoided by using larger straps to secure both the foot and leg within the modified Lidcombe template, thus minimizing accessory motions.

The application of the force by each participant may also be considered a limitation in measuring passive ankle dorsiflexion stiffness. An isokinetic dynamometer applies a constant rate of velocity to the limb in order to measure torque output. The rate of force application to the ankle while using the modified Lidcombe template was not constant among participants; this may have influenced the results compared to a method that used an isokinetic dynamometer. The modified Lidcombe template is unable to account for the influence of velocity on joint movements like an isokinetic dynamometer does because the measurements were passive. It also is unable to account for the influence of muscle activation patterns when going through DF ROM and thus does not demonstrate how different muscles influence stiffness when completing a task such as a jump landing. To better assess the role of ankle brace and tape compliance, stiffness should be measured using an isokinetic dynamometer even if it is not necessarily applicable to a clinical or functional setting.
Lastly, prior intra- and interrater reliability results for the modified Lidcombe template were based upon taking a single measurement when a predetermined magnitude of 80.4 N of force was applied to the ankle.\textsuperscript{30} Stiffness coefficients for the present study were calculated using multiple measurements in order to form an exponential model for a regression line. It is possible that the reliability of the modified Lidcombe template changed with the addition of more measurements and varying amounts of force being applied to the ankle joint. Preliminary data to establish the validity of this measurement could have determined that the modified Lidcombe template was the best tool for this study.

The lack of statistical significance for the stiffness coefficient may not be attributable completely to the modified Lidcombe template. The significant increase in errors on the LESS may be related to the evaluator being a novice rater or being biased by not being blind to the condition applied to a participant in a given trial. However, the researcher who rated each trial of the LESS was trained by one of the original creators of the test, and a study evaluating LESS scores between novice and expert raters reported strong reliability between these groups.\textsuperscript{21}

The increase in errors on the LESS may also be attributed to a psychological factor in the participants. We assumed that participants would not knowingly alter their jump-landing biomechanics when external ankle support is applied, but they may have done so unintentionally. Application of a device encircling a single limb for the first time may alter one’s proprioceptive messaging by stimulating mechanoreceptors. This neuromechanical change may result in a decreased ability of the lower extremity to
accommodate to perturbations during a jump landing. This suggests the need for further studies to determine how various applications affect the lower extremity. For example, wrapping a piece of tape around the calf or wearing extra pairs of socks may influence the stimulation of mechanoreceptors and, consequently, affect one’s neuromechanical response. This would help elucidate any effect on landing biomechanics that exists solely because of the pressure of something encircling the leg, ankle, or foot.

**Future Research**

Further research may provide more insight into the decision about the use of prophylactic ankle braces during sports participation. This study is limited only to the application of the basket weave taping technique and ASO ankle brace. Many different types of braces and taping techniques, including moleskin reinforcement or variations of the basketweave, may result in different effects on jump-landing biomechanics and should be investigated further. Many athletes also wear external ankle support on a regular basis and this study was limited to those who do not. Those that wear prophylactic support for every practice or competition may exhibit biomechanical adaptations to account for the restrictions from the brace or tape. Similarly, external ankle supports lose their restrictive properties over time.\textsuperscript{52} Investigators should study the change in jump-landing biomechanics throughout a longer period of activity, for example, following exercise or several weeks of wearing external ankle support.

With the ever-growing number of ACL reconstructed patients, it would be worthwhile to investigate how this study can be applied to those who have already sustained an ACL tear. As a result of more people sustaining ACL injuries, many athletes
that play a sport or position that is considered to be high risk, such as a football offensive lineman, wear prophylactic knee braces. Since wearing a prophylactic ankle brace has an effect on jump-landing biomechanics, there also is the potential for biomechanical changes when wearing prophylactic knee braces. In particular, it would be interesting to investigate one’s jump-landing biomechanics when wearing both a prophylactic knee brace and external ankle support.

**Conclusions and Clinical Relevance**

The results of this study suggest that wearing a prophylactic ankle brace may increase the risk of ACL injury because participants committed a greater number of biomechanical errors during execution of the Landing Error Scoring System. Compared to wearing no external ankle support, ankle taping had no effect on jump-landing biomechanics. The modified Lidcombe template was unable to detect changes in stiffness that accounted for the alterations in jump-landing biomechanics. Nonsignificant results could be due to problems of validity or reliability, compensation at the subtalar joint, inconsistent application of force by the participants, or measuring dorsiflexion only with the knee extended.

Decreased DF while wearing the brace appears to be the main contributing factor to increased errors on the LESS. The most common errors associated with wearing the ASO ankle brace were asymmetrical foot contact, lateral trunk flexion, and foot external rotation. While these positive findings are assumed to be attributable to decreased DF, it is possible that psychological factors in the participants or bias of the LESS evaluator could have influenced the results.
As the evidence about ACL injury risk continues to grow, patients and clinicians will be able to formulate stronger clinical decisions about injury prevention. While chronic ankle instability is extremely common and bracing is a frequently used method to control it, an ACL injury is, comparatively, much more serious. Any decision to wear an ankle brace, in light of the results of this study, should be made with full consideration of an athlete’s welfare and risk exposure. Preseason screening of jump-landing biomechanics using a measurement tool such as the LESS can properly identify those that may already be at risk for ACL injury. In athletes that are already at a risk of ACL injury based on overall LESS scores, we recommend that their use of prophylactic ankle bracing be carefully weighed in order to prevent a possible increase in jump-landing biomechanics. The results of this study provide insight into only one factor that should be considered when making such a decision.
References


Appendix A: Scoring Sheet for the Landing Error Scoring System

<table>
<thead>
<tr>
<th>ID</th>
<th>Rater</th>
<th>Trial</th>
<th>Sex</th>
</tr>
</thead>
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Note: Please review the LESS Instruction Sheet prior to scoring individuals on the LESS. Initial Contact is the frame before foot flat.

1. Knee Flexion at Initial Contact: <30 deg.
   - Error (1)  □  No Error (0)

2. Hip Flexion at Initial Contact: Hips are NOT flexed
   - Error (1)  □  No Error (0)

3. Trunk Flexion at Initial Contact: Trunk is NOT flexed
   - Error (1)  □  No Error (0)

4. Ankle Plantar-Flexion Angle at Initial Contact: Land Heel to Toe (or) Flat foot
   - Error (1)  □  No Error (0)

5. Asymmetrical Foot Contact: NOT Symmetric
   - Error (1)  □  No Error (0)

6. Lateral Trunk Flexion at Initial Contact: Trunk is NOT vertical
   - Error (1)  □  No Error (0)

7. Medial Knee Position at Initial Contact: Knee medial to midfoot
   - Error (1)  □  No Error (0)

8. Stance Width: > shoulder width
   - Error (1)  □  No Error (0)

9. Stance Width: < shoulder width
   - Error (1)  □  No Error (0)

10. Max IR Foot Position: Toes >30 deg. IR
    - Error (1)  □  No Error (0)

11. Max ER Foot Position: Toes >30 deg. ER
    - Error (1)  □  No Error (0)

12. Knee Flexion Displacement: < 45 deg of flexion displacement following initial contact
    - Error (1)  □  No Error (0)

13. Hip Flexion Displacement: Hips DO NOT flex more than at initial contact
    - Error (1)  □  No Error (0)

14. Trunk Flexion Displacement: Trunk DOES NOT flex more than at initial contact
    - Error (1)  □  No Error (0)

15. Maximum Medial Knee Position: ≥ great toe
    - Error (1)  □  No Error (0)

16. Joint Displacement: Sagittal plane
    - Soft (0)  □  Average (1)
    □  Stiff (2)

17. Overall Impression
    - Excellent (0)  □  Average (1)
    □  Poor (2)
