

THE EFFECT OF LOAD PLACEMENT ON LOWER EXTREMITY JOINT BIOMECHANICS IN LANDING

by

Joseph Riley Patteson

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by

Joseph Patteson

Greenville, NC

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Approved by:

Faculty Mentor (signature required):



Patrick M. Rider

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requirements for the degree of

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Approved by \_\_\_\_\_

Chairperson of Supervisory Committee

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EAST CAROLINA UNIVERSITY

ABSTRACT

The Effect of Load Placement on Lower Extremity Joint Biomechanics in Landing

by Joseph Patteson

Chairperson of the Supervisory Committee: Patrick Rider  
Kinesiology Department

Landing is a critical component to athletic performance. Many studies have investigated the effect that added mass has on landing biomechanics finding when load is added to the body during a landing task, an increase in both vertical and anterior-posterior ground reaction forces is observed. These studies have focused primarily on adding mass to the trunk segment. It is unclear, however, if these adaptations would persist if mass was added to other body segments. We hypothesize that altering the load placement on the body will alter the kinetics and kinematics of lower extremity joints during landing from a vertical drop jump. The purpose of this study is to determine the effect of load placement on joint biomechanics during jump landing compared to unloaded jump landing. 7 healthy college-aged individuals participated in this study. Participants were outfitted for bilateral lower-extremity 3D motion capture. Vertical ground reaction forces (VGRF) were collected from two embedded force platforms. Each participant completed 3 successful jump landings in 3 conditions (unloaded, loaded thigh, and loaded trunk). 10% of the participant's mass was added to the trunk via a weighted vest and 5% was added to each thigh via weighted, form fitting pants during the separate loaded conditions. Landing height was standardized to 50 ( $\pm 2.5$ ) cm. Student's paired samples T-tests were used to detect any mean differences with ( $p < 0.05$ ) indicating significance. A significant reduction ( $p = .004$ )

in peak VGRF was found between the loaded thigh condition ( $1935.4\text{N} \pm 419.9\text{N}$ ) and the loaded trunk condition ( $1755.7\text{N} \pm 308.9\text{N}$ ). Kinematic results showed that during the loaded thigh condition, participants landed with more extended knees ( $-12.3 \pm 5.9$  deg) than the unloaded ( $-13.1 \pm 4.6$  deg) and loaded trunk ( $-15.0 \pm 6.5$  deg) conditions however these results were not significant. Hip flexion angle increased in the loaded trunk condition ( $-22.5 \pm 9.3$  deg) compared to loaded thigh ( $-15.3 \pm 7.6$  deg) and unloaded ( $19.6 \pm 6.2$  deg) although these results were not significant. Our results supported our hypothesis that altering the load placement on the body will alter the kinetics and kinematics of the lower extremity joints during landing from a vertical drop jump. Participants generally landed with more hip and knee extension during thigh loaded conditions and more hip flexion when loaded at the trunk, signifying altered joint kinematics due to load placement.

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## INTRODUCTION

Anterior Cruciate Ligament injuries are debilitating and costly injuries to athletes at all levels of sport (Gottlob, et al., 1999), and with over 200,000 ACL injuries each year in the United States (Miyasaka et al., 1991) this injury is a major concern for athletes all across the nation. Studies have shown that 70% of ACL injuries are considered non-contact injuries, meaning there is a lack of contact with another player. Furthermore, most of these non-contact ACL injuries occur during athletic movements in which the athlete is negatively accelerating such as in jump-landing tasks (Agel et al., 2005). The non-contact nature of these injuries suggests that the mechanism of injury can be attributed to natural internal forces created in response to the external stimuli rather than external forces alone.

In any landing tasks, the immediate force loaded on the body is controlled by the ground reaction force and the flexion angles of the hip, knee, and ankle joints (Yeow et al., 2010). Excessive force applied to the knee in particular can result in injury to the stabilizing ligaments of the knee. The total force applied to the knee ligaments during landing is “determined by the balance of muscle forces, ground reaction forces (GRFS), and joint-contact forces applied to the lower leg” (Pflum et al., 2004). During flexion of the knee, there is co-activation of the quadriceps and hamstrings which serve to stabilize the knee through the motion. This activation of the quadriceps, or the knee extensors, serves to “arrest the downward motion of the body” through an eccentric contraction which dissipates the ground reaction forces (Palmieri-Smith et al., 2007). These findings suggest that increased flexion in the knees during any landing would protect the knees

from injury. Another study reviewed 71 non-contact ACL injuries and found that most of them occurred while the knee was close to full extension (Boden et al, 2000). Two other studies supported these findings and specifically stated that the injuries occurred at less than 30 degrees of flexion (Olsen et al., 2004; Cochrane et al., 2007). Thus landing with extended knees is a prominent risk factor for ACL injury.

There are a few explanations as to why stiff landings put people at such a high risk for ACL injury. A reasonable theory is that the anatomy of the tibial plateau has a posterior slope, which creates an anterior shear force at the knee when the load from the body is applied directly down on an un-flexed knee (Hashemi et al., 2008; Hashemi et al., 2010). This anterior shear force heavily strains the ACL and if it is applied in excess amounts can injure the knee. It is also known that quadriceps activation in an extended position can create tibial shear force, which would add to the shear force already created via the morphology of the knee (DeMorat et al., 2004).

Many studies have investigated the effect that added mass has on landing biomechanics. Studies have shown that when load is added to the trunk during a landing task, an increase in both vertical and antero-posterior ground reaction forces are observed which would increase the force experienced at the ACL. Additionally, studies have shown that an added trunk load particularly increases the demands on the hip, knee, and ankle extensor/plantarflexors and alters landing techniques at each of these joints (Kulas et al, 2008).

Research so far has focused primarily on adding mass to the trunk segment. In the variety of padded sports played today, additional loads are not always solely placed on the trunk. It is unclear if these adaptations would be consistent if mass was added to other body

segments. The purpose of this study was to review how altering load placement on the body will affect the biomechanical strategies of landing from a vertical drop jump in order to better understand the effects of load on the biomechanics of landing. Our hypothesis was that altering the load placement on the body will alter the kinetics and kinematics of the lower extremity joints during landing from a vertical drop jump.

## REVIEW OF LITERATURE

The overall purpose of the study is to see how knee extensor strength correlates to knee flexion during landing from an athletic jump. The specific purpose of this project is to see how varying weight distribution will alter the demands on the knee extensors, and in turn how these altered demands will affect knee flexion during landings. This literature review will examine knee joint landing biomechanics and is organized as follows: 1) Importance of studying landing knee mechanics, 2) Mechanics of human landing, 3) The Danger of extended knees, 4) Quadriceps strength and knee angle during landing.

### *Importance of Studying Landing Knee Mechanics:*

There has been a vast amount of research done that studies how people land, and with the high prevalence of ACL injuries amongst high school and college athletes, a lot of this research has been focused on ACL injuries. In an article published in 1997 in *The Journal of Bone and Joint Function Inc*, it was stated that there are approximately 95,000 ACL injuries occur each year (Frank and Jackson, 1997); plus with increasing size and participation of athletics in both the school and recreation settings since 1997, it can only be assumed that this number has increased in the past 15 years. ACL injuries are both debilitating and costly to athletes (Gottlob et al., 1999), especially when one considers medical costs and recovery time. The even more concerning part of the issue is that roughly 70% of ACL injuries are deemed to be non-contact injuries (Agel et al., 2005), meaning that “mechanism of injury is unrelated to contact with another player or object” (Cruz et al, 2013). Most of noncontact ACL injuries occur during athletic movements in

which the athlete is negatively accelerating such as in jump-landing or pivoting (Agel et al., 2005). This fact is concerning because it suggests that these ACL injuries can be attributed more to the kinematics and kinetics naturally occurring in the leg during the athletes' movements, than they are from external forces influencing the knee. Studying the natural biomechanical process of the knee during these negative acceleration tasks can have a huge impact on the way we train and prepare athletes to perform such tasks. Our study will seek to explore the negative acceleration task of landing, and how several different variables and conditions affect landing.

### *Mechanics of Human Landing*

Before one can accurately identify the mechanisms of ACL injury during landing tasks they must first understand the mechanics, kinetics, and kinematics occurring in the knee during landings which result in non-contact ACL injuries. In any landing task the impulse felt by the body is controlled by the ground reaction force and the flexion angles of the hip, knee, and ankle joints (Lee et al., 2010). If the external ground reaction force is not matched by the internal forces created by the musculoskeletal system, the body will experience some form of injury as result (Fu et al. 2007). For our study we want to focus in on the forces being applied to the knee.

The total force felt by the knee ligaments during a landing task is “determined by the balance of muscle forces, ground reaction forces, and joint-contact forces applied to the lower leg” (Pflum et al. 2004). The knee model of ACL forces during a drop landing presented by Pflum and colleagues, suggest that three factors contribute to the total shear forces felt by the lower leg: 1) the anterior shear force supplied by the patellar tendon; 2)

the anterior shear force from the compression of the tibiofemoral joint; and 3) the posterior shear force applied by the ground reaction force (2004). The summations of these three forces supply the total anterior -posterior force in the knee during landing.

Since the ACL functions to limit the anterior movement of the tibia, it is only natural that an excessive amount of anterior shear force experienced at the knee would create stress on the ACL, and possibly strain or even rupture the ligament. The body actively acts to reduce these shear forces through several different mechanisms, such as increasing flexion in the knees. During any flexion in the knee the leg will experience passive contraction of the knee flexors, i.e. the hamstrings, along with eccentric contraction in the knee extensor muscles, the quadriceps. This activation of the quadriceps during flexion actively serves to “arrest the downward motion of the body”, and the eccentric force produced by the quadriceps is an efficient way of dissipating the ground reaction forces (Palmieri-Smith et al., 2007). The passive concentric contraction of the hamstrings along with the active eccentric contraction of the quadriceps is known as a co-contraction and serves to stabilize the knee during active flexion. Once the greatest flexion angle has been reached, the quadriceps and the hamstrings reverse their action, the quadriceps actively contract concentrically and the hamstrings eccentrically contract, to cause the knee to extend and return to anatomical zero. At this lowest point, the ground reaction force and the downward force from the body are equal meaning the body’s center of mass comes to a brief stop. The concentric quadriceps contractions then cause the ground reaction forces to exceed that of the downward force of the body, causing the knees to extend and the center of mass to return to in natural position.

### *The Danger of Extended Knees*

Boden et al, published a study in June of 2000, titled *Mechanisms of Anterior Cruciate Ligament Injury*, which reviewed 100 cases of ACL injury; 71 of which were non-contact injuries. This study found that most of the non-contact injuries occurred while the knee was close to full extension and there was excessive internal rotation of the tibia (Boden et al. 2000). Further, video reviews of ACL injuries that have happened during game play in team handball and football have found that most all of the injuries happened whilst the knee was in less than 30 degrees of flexion (Olsen et al., 2004; Cochrane et al., 2007). Studies such as these, suggesting that landing with extended knees is a risk for ACL injury, have prompted the need to understand why landing with extended knees puts people in such a high exposure for injury.

The first explanation as to why landing with extended knees is such a risk lies in the anatomy of the knee. The articular surfaces of the tibiofemoral joint and the three-dimensional geometry of the femoral condyles interacting with the tibial plateau during any landing have a direct impact on the mechanical forces occurring in the knee. Hashemi and colleagues have published multiple articles reviewing the medial and lateral tibial slopes of the knee and how these tibial slopes affect the biomechanics of the knee. The general findings were that though these tibial slopes can vary between subjects and genders, both the medial and lateral tibial plateaus tended to have a posterior slope (Hashemi et al., 2008; Hashemi et al., 2010). Naturally, when the femoral condyles land on the tibial shelf with all the weight of the body behind them, it tends to push the lower leg in an anterior direction; thus creating an anterior shear force that strains the ACL. However, this only comes into effect when the subject is landing with an extended knee, otherwise the posterior slope of the tibial plateau would be horizontal or a zero slope if

the knee is flexed. Hashemi and his colleagues identify this biomechanical stressor on the knee as a significant non-modifiable risk factor in their publication titled *Shallow Medial Tibial Plateau and Steep Medial and Lateral Tibial Slopes* (2010). McLean and his colleagues later studied how this knee joint morphology can act as a predictor of ACL injury risk during an *in vivo* dynamic landing. Their study further confirmed the findings of Hashemi et al, in that an individual's knee anatomy is directly associated with their risk during landing (McLean et al., 2010).

A second major theory used to explain why an extended knee is a dangerous position for the ACL is related to the knee extensors acting as stressor on the ACL. However, this particular topic is slightly more controversial than the knee morphology theory in the debate of ACL risk factors. First, it is widely accepted that in a slightly flexed position the angle between the patella tendon and the tibial shaft actually cause a tibial shear force when the quadriceps' are activated (Demorat et al, 2004). More specifically, it has been found that the quadriceps strain the ACL the most between 0 and 30 degrees of flexion; a fact that is even more compelling considering the hamstrings are less able to protect the ACL with increasing extension (Pandy and Shelburne, 1997). This, however, is where the theory splits into two countering arguments.

Some studies have suggested that quadriceps loading alone can cause enough strain in the ACL to cause injury. Demorat et al (2004) found that at 20 degrees of flexion, and quadriceps load of 4500 Newtons is sufficient to cause injury to the ACL in some cases. They then justified this force capacity for the quadriceps using isometric max testing on each of the subjects. This assertion seems to be supported by the studies mentioned in the last paragraph stating that quadriceps create the most strain on the ACL

when the knee is close to full extension (Pandy and Shelburne, 1997). Also, in response to counter arguments, it is pointed out that during landing the quadriceps are contracting eccentrically, which is known to have a larger max than the concentric isometric max obtained in the DeMorat study.

In more recent literature it has been shown that the theory DeMorat et al (2004) presented is in fact flawed. Hashemi and his colleagues performed an *in vitro* simulation of a jump-landing to determine how quadriceps pre-activation forces affected ACL strain. The study found that not only does that pre-activation of the quadriceps cause no additional ACL strain, but also that with increasing levels of quadriceps preactivation they serve to have protective effects for the ACL (Hashemi et al, 2010). Furthermore, a study even more recently has found that peak ACL strain is actually reached prior to contact with the ground (Taylor et al., 2011); solidifying the stance that even though quadriceps do strain the ACL, they are not solely responsible for any ACL injury. Then to further disprove DeMorat's proposed theory, a second simulation study by Domire et al, found that the 4500 N of quadriceps force cited in DeMorat's 2004 study is not realistically attainable in the 50 ms window of time from foot contact during which most ACL tears occur (Domire et al., 2011). These more recent findings actually serve to suggest increasing quadriceps strength and activation is beneficial to the knee, and will help to protect the knee in the event of landing in an extended position.

### *Quadriceps Strength and Knee Flexion*

For the purposes of our study we will be reviewing quadriceps strength and assessing not how it affects the ACL strain at landing, but rather the correlation the extensor strength has with knee flexion at landing. The premise behind our research is

that, since quadriceps force alone is unlikely to be responsible for ACL injury, increasing quadriceps strength will act as a positive protective force for the knees. As stated earlier, the legs will increase knee flexion during landings as a mean of controlling and absorbing the impact from the ground reaction force (Lee et al, 2010). Furthermore, as part of this increased flexion, there is a greater demand on the knee extensors. Devita et al, found that landing with stiff knees decreased the direct demand on the knee extensors, and actually decreased the total energy absorbed by the body by 19%, when compared with fluid landings (DeVita and Skelly, 1992). Given the roll that the extensors play in dissipating the energy during landings (Zhang et al., 2000), it seems reasonable to suggest that a stiff landing (landing with extended knees) is a means of compensating for knee extensor weakness. Thus, greater quadriceps strength will result in greater knee flexion and increased protection of the ACL during landings.

What will separate our study from other studies is that we will be examining the relationship between knee extensor strength and knee angle at initial contact with the ground, and not necessarily tracing the movement through the landing. A different study in 2004 reviewed the relationship between knee extensor strength and maximum knee flexion during landings (Salci et al., 2004), however as previously stated ACL injury at its greatest risk when the knee is at less the 30 degrees of flexion (Boden et al., 2000; Olsen et al., 2004; Cochrane et al., 2007). Since, Salci's study reviewed maximum knee flexion the data is likely not as helpful in regards to ACL injury statistics as it would be had the study reviewed initial knee flexion. Krosshaug and colleagues reported findings in video analysis of 39 basketball ACL injuries that each of the injuries happened within 50 milliseconds of initial ground contact (Krosshaug et al., 2007). In Salci's study the

maximum knee flexions being recorded was not reached within this 50 millisecond window and therefore further suggest that any findings from the study are likely not related to ACL injury risks. Our study will measure knee extensor strength compared to knee flexion angle at initial contact during landings, which will be within the 50 millisecond window for injury and will allow us to assess the preferred angle with which people land. We believe this will provide use with data that is useful in gaining insight into the role that quadriceps strength plays in protecting the ACL during landing tasks.

Also, during our study we plan to further test how multiple other parameters, such as fatigue, weight distribution and loading, and active distraction, affect this correlation between strength and knee flexion during landings. There have been separate studies that have reviewed each of these factors and how they relate to different landing tasks. The purpose of doing this in our study is to simulate realistic landing conditions. For example towards the end of a basketball game a person will be moderately fatigued and is not ever going to be entirely focused on how they are landing. Also, we would like to vary the weigh distribution to see if it is possible to identify whether or not people with certain body styles are at greater risk than others.

Fatigue has actually been a pretty heavily researched subject in any study pertaining to musculoskeletal function. In the past fatigue has been difficult to study because it is a very tough thing to quantify and standardize between separate individuals. It has only been since the turn of the century the researchers have begun to see how fatigue affects functional performance. One study in 2006, found that after fatiguing subjects tended to land with a more extended knee, as a means of reducing the load on the extensors (Augustsson et al., 2006). This, again, fits the theory mentioned earlier that

landing with extended knees is a means of compensating for knee extensor weakness. As the muscle fatigues the knee tends to extend more and more while landing to lighten the load on the extensor muscles, which in fact puts them at a greater risk for ACL injury. In our study we wish to duplicate these findings, except for using a dynamic landing task using both legs.

We also wish to assess whether there are any kinematic changes in the knee during landing tasks with the subject being distracted. Research done in 1996, found that the level of attention a subject invests in certain landing tasks can influence the movement strategies employed for the landing (Mulder et al., 1996). These findings were further supported when McNair and colleagues found that when participants devoted higher levels of attention to landing technique they were actually able to decrease the landing ground reaction forces (McNair et al., 2000). Thus, inattention to landing could in fact lead to altered kinematics which results in greater ground reaction forces. Finally, in a study done by Dempsey and colleagues (2013) on the effects of added load, exercise and distraction on landing forces, it was found that distracted drop landings had significantly higher ground reaction forces than any other landing trials (Dempsey et al., 2013). We believe that having the subject aim for a particular target during their jump-landing tasks will distract the subject enough to keep them from focusing on their landing.

Lastly our study will seek to assess how varying weight distribution on the body will affect knee flexion during landing, thus allowing us to calibrate the added stress placed on the knees. Previous studies have found that added load no matter where in the body typically give an increase in both vertical and anteroposterior ground reaction forces

(Lloyd et al., 2000; Tilbury-Davis et al., 1999). More recently a study in 2008 found that a 10% increase in body mass carried in the trunk increased knee extensor moments and work by roughly 25% (Kulas et al., 2008). The question however lies with how will trunk loading alterations compare with hip and thigh loading alterations. My personal hypothesis is that when the hips and thighs are loaded it will result in a greater knee extensor moment, thus causing the subject to tend to land with more extended knees, putting them at greater risk for ACL injury. The trunk tends to stay directly over the knees and thus the knee extensor moment would not change as much when the load is added in the trunk. The results of our study should give data that help to answer this question, and thus give us further information as to who is at greater risk for ACL injury.

In reading this literature review I hope that you have gained a pretty good idea as to what kind of published research is available on the topic. Given the prevalence of ACL injury, there really is an incredibly vast amount of research being done on the topic. However, even with all the current research, there are still holes in the data. The ultimate goal is identify the risk factors the lead to ACL injury and help to find ways of preventing them in the future. We believe that our research is geared towards doing exactly that. There are no published findings of how knee extensor strength correlates to knee flexion at contact during landings. We believe that our findings from this research have the potential to be used in creating quadriceps training programs that will help to prevent future ACL injuries.

## METHODOLOGY

### Subject Selection:

Ten recreationally active subjects participated in this study, that met the following inclusion criteria for this project were: a) between the ages of 18 to 30 years old, b) healthy and free of any skeletal, nervous, and mental impairments, c) BMI between 19.5 and 28 kg/m<sup>2</sup>, d) and have been at least recreationally active for one year prior to participation in the study. Testing protocols were approved by the University Institutional Review Board for human research.

### Subject Recruitment

Subjects were recruited for this study primarily through two different methods. The first method was simply word of mouth. The research team for this study was pretty extensive, including 6 members, and each of the members reached out to friends and family who fit the inclusion criteria and inquired about their willingness to participate in the study. The second major method of recruitment was in class recruitment. In class recruitment consisted of a researcher visiting several classes and giving a short summary of the purpose, protocol, inclusion criteria, and perks of participating in the study. Sign-up sheets were then passed out through the class and students who signed-up were reached out to via email. Each of the students who met all the inclusion criteria, were sent a follow-up email, and scheduled to bring into lab for testing.

## Testing Protocol

On the first day of testing participants were asked to read and sign an informed consent form (Appendix A), as well as confirm the demographic information qualifying them for the study (Appendix B). Participants were then measured for height and weight which were used to calculate official BMI scores. An in depth summary of the 2-day protocol was reviewed for the participants describing the equipment, procedures, and expectations of them during the study. Finally, participants were given the opportunity to view and familiarize themselves with the equipment that was to be used for the study.

On the second day of testing, upon entering lab, participants were asked to change into form-fitting spandex shorts. Participants were then outfitted for bilateral lower-extremity 3D motion capture. The 3D motion capture data was collected using Qualisys Track Manager (QTM) software and an 8 Pro-Reflex camera system collecting at 120 Hz. Reflective markers used for the motion capture were placed as follows:

- a. Foot plate
- b. 1<sup>st</sup> and 5<sup>th</sup> metatarsal heads- removed after static calibration
- c. Lateral and Medial ankle- removed after static calibration
- d. Heel
- e. Shank plate
- f. Lateral and Medial knee-removed after static calibration

g. Thigh plate

h. R and L greater trochanters- removed after static calibration

i. R and L iliac crests –removed after static calibration

j. R and L PSIS

k. R and L ASIS

l. Sternum

m. R and L Acromion

n. Back along spine between scapula

Prior to each having the participant jump in each condition, a static calibration trial was collected. This trial would later be used to build a model of the subject.

Each participant completed 3 successful jump-landings in 3 separate conditions. The unloaded condition involved the participant completing the jump task with no additional load placed on the body. The loaded legs condition involved the participant completing the jump task with 5% of their total mass added to each of the participant's thighs using specially made form fitting pants. The loaded trunk condition involved participants completing the jump task with 10% of their total mass added to their trunk via a weighted vest with the weights spread evenly between the front and the back of the vest.

Participants started each jump task, jumping off a 40cm box. A jump task was considered successful when fall length, tracked by the vertical position of the PSIS markers, was  $50(\pm 2.5)$  cm and both feet hit on separate embedded AMTI force platforms (AMTI Model LG-6). The force platforms were used to collect the force data which would later be combined with the kinematic data from the motion capture system to calculate the kinetics occurring in the lower extremities during each landing. The order in which each condition was completed was randomly assigned in order to account for any learning affects or adaptations to landing strategy as the subjects moved through the protocol.

Participants were instructed to touch a target, placed just above their heads, during each jump task. The target served help participants jump in the appropriate height range, and also to make the landings more natural.

### Data Analysis

Markers were tracked, labeled, and gap-filled using Qualisys Track Manager (QTM) software (Qualisys, Gothenburg, Sweden). In V3D, a model was built and inverse dynamics were used to calculate the specific joint torques and kinematics at each of the lower extremity joints. In the loaded conditions the model was adjusted for the added load by altering the segment inertial properties to account for the specific load conditions. Thus, in the loaded trunk condition, an additional 10% of mass is added to the trunk model, and in the loaded legs condition 5% of the subjects mass is added to each thigh.

Once model was built in V3D, we processed and filtered the data using bidirectional low pass Butterworth filter operating a 6 Htz for kinematic marker data and 45 Htz for the force plate data. Initial contact for the landing was indicated as any point at which the force threshold exceeded 5 Newtons on either force plate. Peak knee flexion signified the end of the landing movement. Peak torque and peak GRF events were calculated between this initial contact point and the end of the landing movement. We then isolated the data so we were only viewing joint positions at initial contact, peak vertical GRF, and peak joint torques for each of the landings we recorded. These specific events were then converted into event metrics and exported into Microsoft Excel for statistical analysis.

### Statistical Analysis

For each individual subject, average flexion values and GRFs between each of the legs were calculated for each trial of each condition. Averages of the right and left joint values, to find the mean flexion angles of both the joints were computed. Finally, to find the general trends, all of the subject's values at each of the joints is averaged and graphed comparing unloaded to each loaded condition. Student's paired samples T-tests were used to detect any differences between these mean values with ( $p < 0.05$ ) indicating significance.

# RESULTS

## Ground Reaction Forces

Average peak vertical ground reaction forces (vGRFs) increased in both the loaded legs and the loaded trunk condition. In the loaded legs condition there was a significant increase in average peak vGRFs of 13.57% when compared to the unloaded

**Figure 1:**

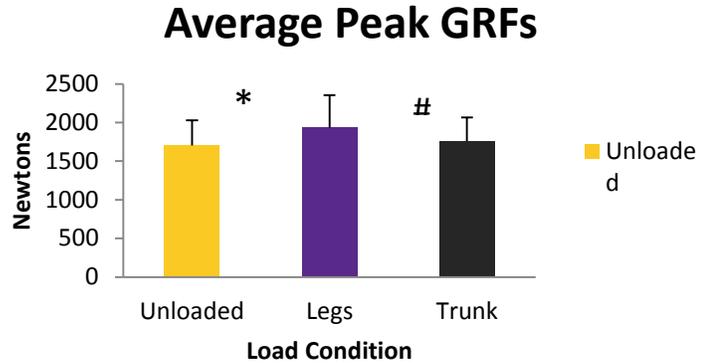


Figure 1: Depicts the average vertical GRFs in each load condition. \* Indicates significant difference between loaded legs and unloaded conditions. # indicates significance between legs and trunk conditions

condition. The loaded trunk condition showed an average increase of only 3.03%, which proved to be insignificant. Interestingly, in 4 of the 10 recorded cases, the average peak vGRFs actually decreased from the unloaded condition to the trunk loaded condition.

**Figure 2:**

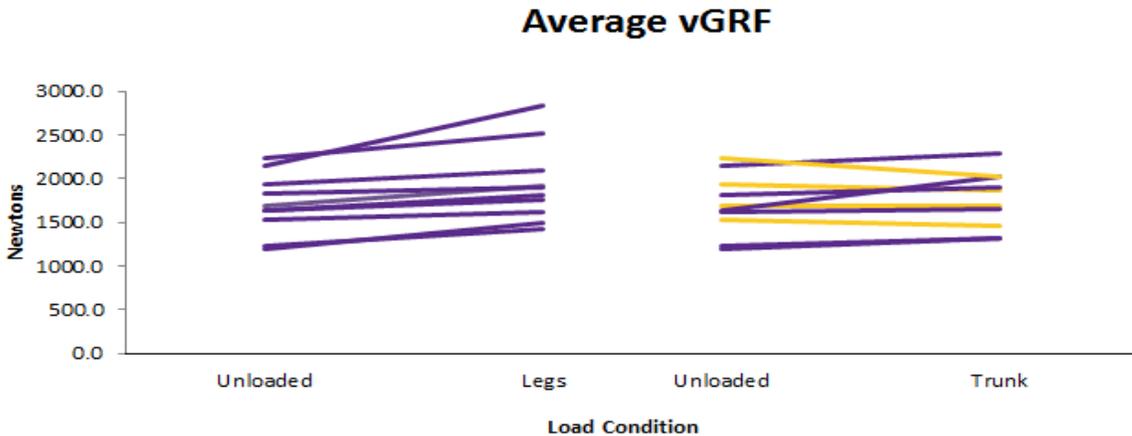


Figure 2: Depicts the changes in average vertical ground reaction forces between the two loaded conditions and the unloaded condition. Purple line indicates an increase in vGRFs; Gold line indicates decrease in vGRF.

There is also a significant difference between the two loaded conditions with the loaded trunk having smaller peak GRFs than loaded legs.

### Joint Kinematic Alterations

Joint flexion angles are reported as a mean of the three trials in each of the load conditions. Each of the flexion angles has been normalized to the subject's natural standing posture, collected during the standing calibration prior to performing the jump task of each condition, rather than from anatomical zero.

Ankle plantarflexion averaged at 31.2 degrees in the unloaded condition. In both the loaded legs and the loaded trunk condition these flexion angles increased (33.2 and 32.4 degrees of flexion, respectively), however this change was insignificant (Figure 3).

Knee flexion angles changed differently depending on the load condition; however none of the changes were significant (Figure 4). In the unloaded condition the average initial knee flexion was 13.1 degrees.

Figure 3:

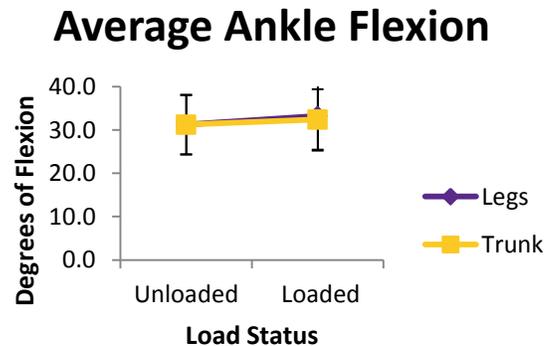


Figure 3: Depicts the change in average ankle flexion between load conditions compared to the unloaded condition.

Figure 4:

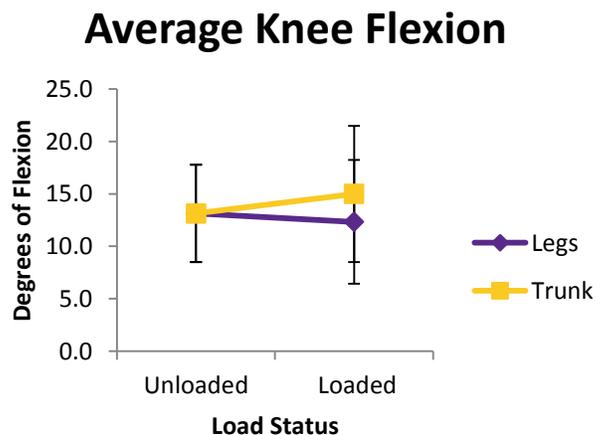


Figure 4: Depicts changes in average ankle flexion between the load conditions compared to the unloaded condition

When loaded at the legs average knee flexion decreased by almost a full degree, moving to an average of 12.3 degrees of flexion. When loaded at the trunk, knee flexion increased moved to an average of 15 degrees of flexion.

Average hip flexion for the unloaded condition was 19.6 degrees. Participants generally decreased hip flexion when loaded at the legs, and increased hip flexion when loaded at the trunk (Figure 5). Average hip flexion when loaded at the legs was 15.3 degrees of flexion. Then when loaded at the trunk the average hip flexion was 22.5 degrees of flexion. Neither of the load conditions showed a significant

difference when compared to the unloaded condition. However, when comparing the two loaded conditions to one another, there is a significant difference in hip flexion.

When loaded at the trunk, the alterations seen at the hip were not uniform in all subjects. 3 of the 10 subjects tested actually decreased their hip flexion in response to the trunk load (Table 1). This is completely opposite the other 7 subjects who increased their hip flexion with the trunk load. When you separate the groups into a decreased flexion group and an increased flexion group, the both groups actually saw a significant difference in hip angle from the unloaded condition.

Figure 5:

### Average Hip Flexion

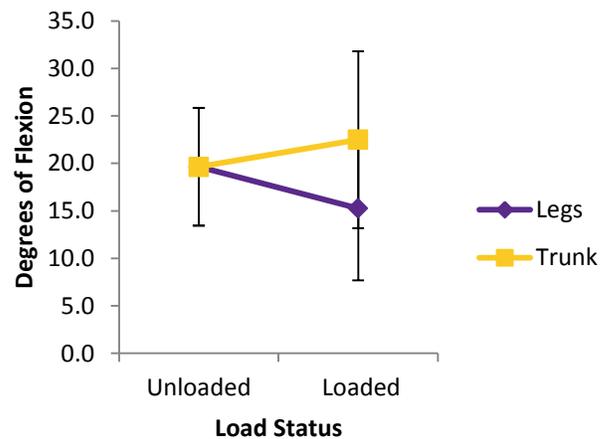


Figure 5: Depicts the changes in average hip flexion between the load conditions as compared to the unloaded condition.

**Table 1:**

	Hip Flexion Angle (degrees)		
	Unloaded	Legs	Trunk
S008	-24.2	-27.1	-32.2
S009	-12.4	-22.6	-23.0
S010	-19.8	-11.2	-28.6
S011	-12.8	-6.5	-15.4
S012	-24.0	-10.9	-24.9
S013	-32.58	-16.50	-19.21
S014	-23.76	-23.36	-33.52
S015	-17.50	-2.22	-6.26
S016	-16.38	-20.10	-9.14
S017	-13.0	-12.2	-32.8

Table 1: Represents the average hip flexion values at initial contact in each of the landing conditions for each of the subjects. Highlighted subjects decreased hip flexion from the unloaded to the loaded trunk condition.

### Joint Torque Alterations

The joint torque results that are reported are a mean of the 3 trials from each load condition at each of the joints then averaged between right and left legs to discount any lateral asymmetries in landing tasks.

Ankle torque increased from the unloaded condition in both

**Figure 6:**

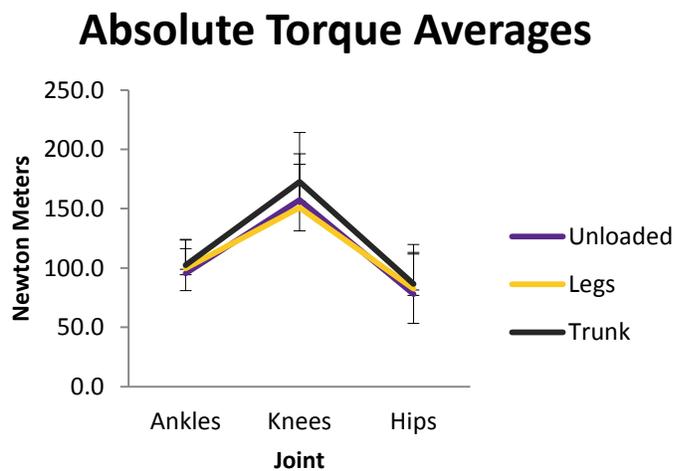


Figure 6: Graph depicting the absolute torque averages. Purple line represents the unloaded condition, gold line represents the leg loaded condition, and black line the trunk loaded condition.

load conditions but these changes were insignificant (Figure 6). The unloaded condition saw an average absolute torque value of 95.3 Newton meters (Nm). In both of the loaded conditions we saw an average ankle torque of 99.8 (leg) and 102.3 (trunk) Nm of torque.

The average knee torque in the unloaded condition was 157.3 Nm. In response to the leg load, the knee torque actually decreased (150.9 Nm), although the difference was not significant. In response to the trunk load there was a significant increase in the average absolute torque, which jumped to 172.8 Nm (Figure 6).

The average absolute hip torque in the unloaded condition was 77.8 Nm. This torque increased in both of the loaded conditions, jumping to 82.4 Nm in the legs loaded condition and 86.5 Nm in the trunk loaded condition (Figure 6) but neither of these increases was significant.

**Table 2:**

Load Condition	Unloaded	Legs	Trunk
	Flexion Angle (degrees)		
Ankle	31.2	33.2	32.4
Knee	13.1	12.3	15.0
Hip	19.6	15.3	22.5
	GRFs (newtons)		
Peak GRFs	1704.1	1935.4	1755.7
	Torque (Newton meters)		
Ankle	95.3	99.8	102.3
Knee	157.3	150.9	172.8
Hip	77.8	82.4	86.5

Table 2: Represents the overall mean values for each of the variables we considered from each subject for this study.

## DISCUSSION

The purpose of this study was to review how altering load placement on the body will affect the biomechanical strategies of landing from a vertical drop jump in order to better understand the effects of load on the biomechanics of landing. Our hypothesis was that altering the load placement on the body will alter the kinetics and kinematics of the lower extremity joints during landing from a vertical drop jump.

Our variables of concern were the angles at initial contact at the ankle, knee, and hip joints, peak ground reaction forces, and peak torque at each joint. Our primary findings showed alterations in landing strategy leading to increased stiffness (decreased flexion) in the leg loaded condition and increased fluidity (increased flexion) in the trunk loaded condition. Our overall hypothesis was supported in that there were alterations in the kinetics and kinematics of landing when altering the load placement on the body.

### **Ground Reaction Forces**

Previous studies have shown that stiff landings actually result in an increase in total energy absorbed by the body compared with fluid landings (Devita and Skelly, 1992). In this study, average peak ground reaction forces increased in both of the load conditions. The GRF's increased more in the legs loaded condition than in the trunk loaded condition, increasing 13% and 3% respectively. The percent increase in the GRFs is interesting because we added a set load of 10% of the subjects mass to their body. However, the GRF's do not increase by a standard 10%. This difference in GRF response

to the load could be a result of an individual's ability to adapt to the placement of the load on the body.

The 3% increase in GRF's in the trunk loaded condition, shows that the participants were efficient at compensating for the additional load. Subjects appear to have maintained, if not increased, their fluidity in landing with the trunk load. This is not the case when considering the leg loaded condition. Subject's GRFs increased by 13%, suggesting that they are in-fact landing more stiffly when loaded at the legs. The increase in GRFs could be a direct result of the reduced role the knee extensors have in this extended position at dissipating the energy created during landings (Zhang et al, 2000).

There was a significant difference in GRF's between the two loaded conditions. Since each of these conditions have the same additional load on the body, the difference in peak GRFs can likely be attributed solely to differences in landing strategies, and thus supports our hypothesis. Altering the placement of the same load from the trunk to the legs elicited a change in landing strategy that significantly increased the vGRFs experienced in the body, showing that not just load, but load placement can significantly change the way a person lands.

### **Joint Kinematic Alterations**

We examined flexion angles at initial foot contact for each of the lower extremity joints. Previous studies reviewing joint kinematics in landing tasks have generally considered the maximum flexion of each of these joints (Kulas et al, 2008; someone

else), however, as it has been shown that injury normally happens within the first 50 milliseconds of landing (Krosshaug et al., 2007). We examined the flexion angles at initial contact to more accurately analyze the conditions that are related to knee injury.

When discussing the kinematic responses to load, one should consider the lower limb joints as a single linked system, rather than considering each joint individually. Subjects landed more stiffly when loaded at the legs than in the unloaded condition, decreasing flexion at both the knee and the hip. Subjects landed more fluidly when loaded at the trunk, increasing flexion at the knee and hip. There is a significant increase in hip flexion when comparing the leg loaded to the trunk loaded condition. The knee also showed increased flexion from leg loaded to trunk loaded, although this increase was not significant. These findings support our overall hypothesis that load placement alters the biomechanics of landing.

In a supplementary observation to our results, we found that hip joint kinematics really told an interesting story when loaded at the trunk because not all subjects employed the same compensation techniques. Of the 10 subjects tested, 3 subjects decreased flexion, while 7 increased flexion. When you separate these two groups, both showed significant changes in flexion from the unloaded condition. There seems to be evidence of two separate landing strategies to compensate for this increased trunk load. It would be interesting to see if one of the strategies would emerge as dominate with an increasing number of subjects. It is possible that we simply have three outliers who are having a large effect on our data because of our small sample size. Additionally, previous studies have found that trunk load would affect primarily hip

mechanics in drop landings (Kulas et al, 2008). Thus far, we have not considered trunk position in our work but it is possible trunk position could be responsible for these discrepancies at the hip.

### Joint Torque Alterations

It is important to consider joint torques in our analysis of the effect that load placement has on kinetic and kinematic alterations in landing. In any drop landing, the vGRFs induce an external joint flexor torque at each of the lower extremity joints (Sell et al., 2010). The joint torque is a product of the vGRF and the moment arm, which is the perpendicular distance from the applied force (i.e. vGRF) to the point of rotation (i.e. the joint center) (Figure 7). To counterbalance and control the joint flexor torque the body has to produce an internal joint extensor torque (Sell et al., 2010). At individual joints, such as the knee, it is the job of the extensor muscles to activate and create the internal torque to match the external torque. At the knee this torque is created by quadriceps muscle activation via an eccentric contraction. As the quadriceps are activated, they increase ACL strain producing an anterior shear force on the tibia (Yu and Garret., 2007). Increased ACL strain puts the subject at a higher risk for injury, so any method of

Figure 7:

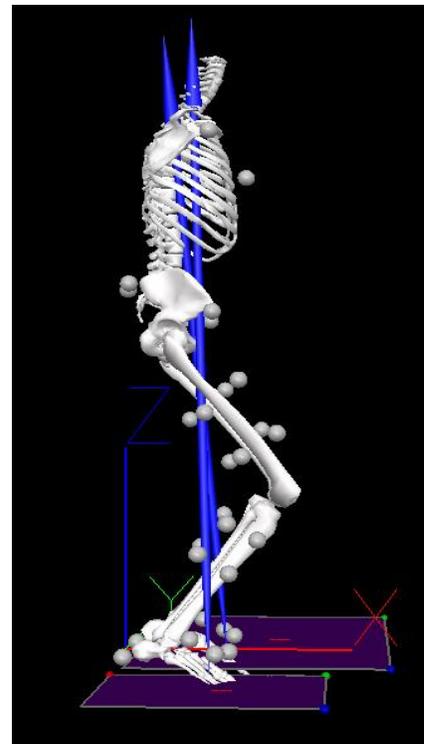


Figure 7: Image create by V3D software depicting the vGRF (blue vectors) tracing through a simulated model at initial contact during a jump landing task..

reducing this shear force is beneficial to the subject. In order to reduce this strain, it would make sense to reduce the torque placed on the knee, via reducing the moment arm. The way to reduce this moment arm would be to keep the center of mass as close to each of

the joint centers as possible, since the vGRF is going to straight through the center of mass (Figure 7). In the unloaded and trunk loaded conditions, the center of mass is close to centered on the trunk. However, when loaded at the legs, the center of mass is lowered towards the pelvis, and is moved away from the center of the lower extremity joints with increased joint flexion.

As shown in the kinematic results section our subjects landed more stiffly in the leg loaded condition. Absolute joint flexor torques at the knee decreased in the leg loaded condition, and

**Figure 8:**

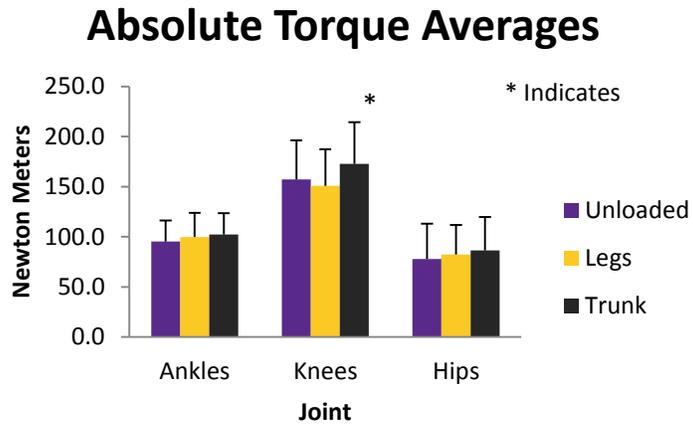


Figure 8: Graph depicting the absolute torque averages. \* Indicates significant difference when compared to the unloaded condition.

**Figure 9:**

### Torque Averages Normalized to Mass

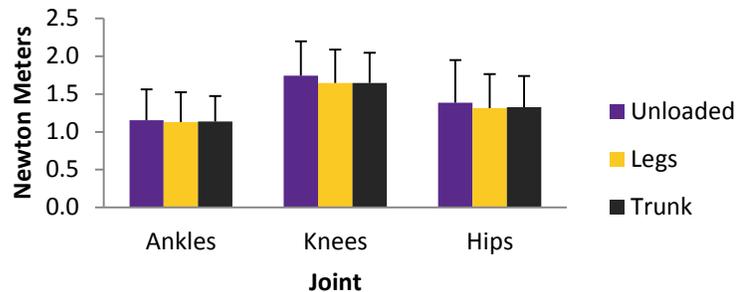


Figure 9: Shows joint torque averages for each of the lower extremity joints after being normalized to each subjects mass

slightly increased at the hip and ankle (Figure 8). When the joint torques were normalized to mass, there is no change in joint torque with the load (Figure 9). The trunk loaded landings were generally more fluid and saw increase initial flexion at each of the joints. Absolute joint torque increased at each joint in the trunk load condition (Figure 8). Again when normalized to mass however, there was no significant change in joint torque (Figure 9). The absolute joint flexion torque values increased solely due to the added mass and the resultant increase in GRFs in both of the load conditions. There were no significant changes in torque production; however, there is a noticeable change in kinematic responses. This may indicate that with added load there was no real alteration in muscular effort, meaning the difference in GRFs are affecting the kinematic responses, and the added load was dissipated in the form of added compressive forces on the skeletal structure.

## CONCLUSION

The purpose of our study was to review the effects that load have on the biomechanical strategies of landing. Our findings supported our hypothesis that altering load placement on the body would alter the kinetics and kinematics of landing from a vertical drop jump. We were able to identify evidence that suggested biomechanical strategy alterations in landing with altered load placements.

Altering the placement of the same load from the trunk to the legs caused subjects to move from a fluid landing to a stiff landing. When the landings are stiffer, the lower extremity muscles have a smaller role in dissipating the vGRFs. Thus, in the loaded legs condition we saw an increase in vGRFs greater than the 10% of mass that was added. Also, in the loaded trunk condition, the landings were very fluid and the increase in vGRFs was less than the 10% of the mass that we added. This relationship implies a trade-off between joint torque and vGRF. Fluid landings have decreased vGRFs but increased joint torques. Stiff landings result in increased vGRFs and decreased joint torques. Joint torque is directly related with musculature effort. In the fluid landings the flexion increased and musculature effort increased, where as in the stiff landings the flexion decreased and thus the musculature effort decreased.

Landing with extended knees (stiff landing) is highly associated with injury to the lower extremity joints, particularly the ACL in the knee. Studies have reviewed landing mechanics and the effects of load as it pertains to ACL injuries. Our results suggest that the placement of said load actually can have variable effects on landing mechanics and

the forces impacting the ACL. Leg load in particular leads to stiff landings, and thus may lead to higher likelihood of ACL injuries.

The effect of load placement on landing joint biomechanics has broad implications for methodological testing purposes during future studies testing the effect of load, as well as for training purposes in athletes and active professionals in all fields.

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## APPENDIX A

*East Carolina University*



### **Informed Consent to Participate in Research** Information to consider before taking part in research that has no more than minimal risk.

Title of Research Study: The effect of knee extensor strength on landing biomechanics and risk for ACL injury

Principal Investigator: Zac Domire

Institution/Department or Division: Department of Kinesiology

Address: 332 Ward Sports Medicine Building, Greenville NC, 27858

Telephone #: 252-737-4564

Study Sponsor/Funding Source: East Carolina University

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Researchers at East Carolina University (ECU) study problems in society, health problems, environmental problems, behavior problems and the human condition. Our goal is to try to find ways to improve the lives of you and others. To do this, we need the help of volunteers who are willing to take part in research.

#### **Why is this research being done?**

The purpose of this research is to investigate the relationship between knee strength and risk for ACL injury. The decision to take part in this research is yours to make. By doing this research, we hope to learn if there are identifiable risk factors for ACL injury.

#### **Why am I being invited to take part in this research?**

You are being invited to take part in this research because you are a healthy volunteer. If you volunteer to take part in this research, you will be one of about 40 people to do so.

#### **Are there reasons I should not take part in this research?**

You should not take part in this research if: you are under the age of 18 or over the age of 30, your BMI is outside of 19.5-28 kg/m<sup>2</sup>, you have had any previous injuries or surgeries to your legs or back, or you have not been at least recreationally active for at least one year prior to participation in the study.

#### **What other choices do I have if I do not take part in this research?**

You may choose not to participate.

#### **Where is the research going to take place and how long will it last?**

The research procedures will be conducted in the Biomechanics Laboratory, room 332 Ward Sports Medicine building at ECU. You will need to come to the lab three times during the study. The total amount of time you will be asked to volunteer for this study is 4.5 hours.

#### **What will I be asked to do?**

You are being asked to do the following:

During the first visit:

- Provide personal information about your general health and activity level
- Read and sign this informed consent document
- Have your height and weight measured
- Perform isometric strength testing on the quadriceps muscles. Electrical stimulation will be used to verify muscular effort. Participants may experience discomfort and minimal pain. (It may be similar to the feeling of a static electricity shock.)

During the second visit:

- Change into form fitting shorts
- Perform isometric strength testing on the quadriceps muscles. Electrical stimulation will be used to verify muscular effort. Participants may experience discomfort and minimal pain. (It may be similar to the feeling of a static electric shock.)
- Have reflective markers placed on your leg to monitor your movement during testing
- Perform several jumping tasks

During the third visit:

- Change into form fitting shorts
- Perform isometric strength testing on the quadriceps muscles. Electrical stimulation will be used to verify muscular effort. Participants may experience discomfort and minimal pain. (It may be similar to the feeling of a static electric shock.)
- Complete an isokinetic exercise session designed to fatigue the quadriceps muscles. Perform isometric strength testing on the quadriceps muscles. Electrical stimulation will be used to verify muscular effort. Participants may experience discomfort and minimal pain. (It may be similar to the feeling of a static electric shock.)
- Have reflective markers placed on your leg to monitor your movement during testing
- Perform several jumping tasks
- Perform another strength testing session the same as the one described above

### **What possible harms or discomforts might I experience if I take part in the research?**

It has been determined that the risks associated with this research are no more than what you would experience during a moderate bout of strength training.

### **What are the possible benefits I may experience from taking part in this research?**

We do not know if you will get any benefits by taking part in this study. This research might help us learn more about the relationship between knee strength and risk for ACL injury. There may be no personal benefit from your participation but the information gained by doing this research may help others in the future.

### **Will I be paid for taking part in this research?**

We will pay you \$30 for participating in this research, if you participate in all three of the sessions. If you are unable or unwilling to participate in all of the sessions, you will be compensated for the sessions you do complete. You will receive \$10 per session. You will be paid in the form of a gift card. The gift card will be given to you at the completion of all study sessions. If you are unable to pick up your card in person, it will be mailed to you.

### **What will it cost me to take part in this research?**

It will not cost you any money to be part of the research.

### **Who will know that I took part in this research and learn personal information about me?**

To do this research, ECU and the people and organizations listed below may know that you took part in this research and may see information about you that is normally kept private. With your permission, these people may use your private information to do this research:

- Any agency of the federal, state, or local government that regulates human research. This includes the Department of Health and Human Services (DHHS), the North Carolina Department of Health, and the Office for Human Research Protections.
- The University & Medical Center Institutional Review Board (UMCIRB) and its staff, who have responsibility for overseeing your welfare during this research, and other ECU staff who oversee this research.
- Zac Domire, the principle investigator, Paul DeVita, the Lab Director, Patrick Rider, the study coordinator, and the students assigned to this study.

### **How will you keep the information you collect about me secure? How long will you keep it?**

Data files will be kept for 5 years after the study is completed. The investigators will keep your personal data in strict confidence by having your data coded. Instead of your name, you will be identified in the data records with an identity number. Your name and code number will not be identified in any subsequent report of publication. The study investigators will be the only persons who know the code associated with your name and this code as well as your data will be kept in strict confidence.

### **What if I decide I do not want to continue in this research?**

If you decide you no longer want to be in this research after it has already started, you may stop at any time. You will not be penalized or criticized for stopping. You will not lose any benefits that you should normally receive.

### **Who should I contact if I have questions?**

The people conducting this study will be available to answer any questions concerning this research, now or in the future. You may contact the Principal Investigator at 252-737-4564 (work days, between 8am-5pm).

If you have questions about your rights as someone taking part in research, you may call the Office for Human Research Integrity (OHRI) at phone number 252-744-2914 (days, 8:00 am-5:00 pm). If you would like to report a complaint or concern about this research study, you may call the Director of the OHRI, at 252-744-1971.

### **I have decided I want to take part in this research. What should I do now?**

The person obtaining informed consent will ask you to read the following and if you agree, you should sign this form:

- I have read (or had read to me) all of the above information.
- I have had an opportunity to ask questions about things in this research I did not understand and have received satisfactory answers.
- I know that I can stop taking part in this study at any time.

- By signing this informed consent form, I am not giving up any of my rights.
- I have been given a copy of this consent document, and it is mine to keep.

\_\_\_\_\_  
**Participant's Name (PRINT)**

**Signature**

**Date**

**Person Obtaining Informed Consent:** I have conducted the initial informed consent process. I have orally reviewed the contents of the consent document with the person who has signed above, and answered all of the person's questions about the research.

\_\_\_\_\_  
**Person Obtaining Consent (PRINT)**

**Signature**

**Date**

\_\_\_\_\_  
**Principal Investigator (PRINT)**

**Signature**

**Date**

APPENDIX B

**The Effect of Knee Extensor Strength on Landing Biomechanics and Risk for ACL Injury**

**Demographic Data:**

Date \_\_\_\_\_

Name \_\_\_\_\_ Phone Number \_\_\_\_\_

Address \_\_\_\_\_

Date of Birth \_\_\_\_\_

Height \_\_\_\_\_

Weight \_\_\_\_\_

BMI (kg/m<sup>2</sup>) \_\_\_\_\_

**Health Questions:**

Have you been at least recreationally active for the past year? Yes No

Do you have any neurological conditions such as Parkinson's disease or history of stroke? Yes No

Have you previously had surgery on your legs or back? Yes No

Do you have any other medical conditions that were not discussed? Yes No  
If yes, please explain \_\_\_\_\_

\_\_\_\_\_

