UNDERSTANDING THE EFFECTS OF PROGRESSIVE FATIGUE ON IMPACT LANDING FORCE AND KNEE JOINT MECHANICS, DURING THE LANDING PHASE OF CONTINUOUS MAXIMAL VERTICAL JUMPS

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UNDERSTANDING THE EFFECTS OF PROGRESSIVE FATIGUE ON IMPACT LANDING FORCE AND KNEE JOINT MECHANICS, DURING THE LANDING PHASE OF CONTINUOUS MAXIMAL VERTICAL JUMPS

By

Paul Michael Leuty

A Thesis Submitted to the Faculty of Graduate Studies Through the Faculty of Human Kinetics In Partial Fulfillment of the Requirements for The Degree of Master of Human Kinetics at the University of Windsor

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UNDERSTANDING THE EFFECTS OF PROGRESSIVE FATIGUE ON IMPACT LANDING FORCE AND KNEE JOINT MECHANICS, DURING THE LANDING PHASE OF CONTINUOUS MAXIMAL VERTICAL JUMPS

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September 27, 2016
Declaration of Originality

I hereby certify that I am the sole author of this thesis and that no part of this thesis has been published or submitted for publication.

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Abstract

The purpose of the current study was to understand the effects of progressive muscle fatigue on the mechanical behavior of the knee joint throughout maximal double leg vertical jumping. A biomechanical methodology was utilized to examine 28 recreational athletes (14 male, 14 female) who completed continuous maximal vertical jumps in 5 s intervals (12 jumps/min) until fatigue. 3D motion capture was utilized to measure the changing kinematics of the participants’ trunk, hip, knee and ankle joint during jump landing. Two parallel force plates were utilized to record the ground reaction forces of the participants at jump landing. Finally, electromyography was also employed to provide data from the muscular activity of 8 muscles on the participants’ dominant leg. Repeated measures ANOVA (p<0.05) with Tukey’s significant post hoc test were used on any significant main effect of time to fatigue, or interactions between time and sex. Analysis of the kinematic data revealed that as fatigue progresses, knee and ankle joint angle during jump landing significantly decreases in both males and females. Analysis of kinetic data revealed that peak landing impact force, normalized to body mass, significantly increases as participants’ fatigue, even though a significant decrease in jump height is observed. Finally, electromyography data showed that as fatigue progressed throughout the trial, activation actually decreased in both the pre-takeoff and pre-landing phases of the jump for both males and females. This work has helped to fill the current gap in understanding knee joint mechanics associated with progressive muscle fatigue, as well as changes in normalized landing force as fatigue progresses.
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List of Abbreviations

ACL: Anterior Cruciate Ligament
ANOVA: Analysis of Variance
BF-L: Biceps Femoris long head
BF-S: Biceps Femoris short head
BF: Biceps Femoris
CMRR: Common Mode Rejection Ratio
CofG: Centre of Gravity
ED: Effective Duration
EMD: Electromechanical Delay
F/E: Flexion/Extension
GL: Gastrocnemius Lateral
GM: Gastrocnemius Medial
GR: Gracilis
H/Q: Hamstring to Quadriceps ratio
Hz: Hertz
JRS: Joint Rotational Stiffness
Kg: Kilogram
kN: Kilonewton
LCL: Lateral Collateral Ligament
MCL: Medial Collateral Ligament
MU: Motor Unit
MVE: Maximal Voluntary Exertion
N: Newton
PCL: Posterior Cruciate Ligament
PE: Potential Energy
QF: Quadriceps Femoris
RF: Rectus Femoris
sEMG: Surface electromyography
SM: Semimembranosus
ST: Semitendinosus
VI: Vastus Intermedius
VL: Vastus Lateralis
VM: Vastus Medialis
Chapter 1
Introduction

1.1 Background

Knee joint injury is common among adolescent athletes as it is estimated that 2.5 million sports related knee injuries are documented annually by United States emergency departments (Gage et al., 2012). Furthermore, Morgan et al., (2014) recorded that approximately 80% of knee joint injuries occur in a non-contact situation, most frequently occurring during single leg jump landing, or sport tasks involving sidestepping. A significant financial burden is associated with these injuries, with estimated costs to the United States healthcare system to be over $1.5 billion annually (Morgan et al., 2014). Although there are many factors implicated in non-contact knee joint injuries, ultimately, when the forces associated with external demand exceed the strengths of the internal structures of the joint, the risk of injury to these structures increases.

A knee joint is considered injured when normal knee function is compromised chronically, acutely, or through repetitive strain, causing the ligaments functioning to support the knee joint to become unstable, or when impacts on the menisci exceed the capability of shock attenuation (Yeow et al., 2011). Non-contact injuries, as defined by Yeow et al., (2011) occur by the foot making contact with only the ground, and represent 80% of all knee joint injuries (Morgan et al., 2014). Specifically, data indicates that in non-contact knee joint injuries the anterior cruciate ligament is stretched or, torn in 70% of occurrences (Donnelly et al., 2012, Morgan et al., 2014, Tortora & Nielsen, 2013).
Several muscles cross the knee joint producing force, which cause motion about the knee joint. In addition to producing forces for movement, these muscles, along with the ligaments and menisci, provide forces that promote mechanical stability about the knee joint. Specifically, Hewett et al., (2005) stated that co-activation of the hamstrings and quadriceps is the primary mechanism that is used by the neuromuscular system to enhance stability about the knee joint. Moreover, work by Delp et al., (1990) indicated varied force contribution from the leg muscles such that, the vastus lateralis and the semimembranosus have the greatest maximal force capacity about the knee joint. This finding was then expanded on by Derouin and Potvin (2005) that the vastus lateralis and semimembranosus have the greatest potential to stabilize the knee joint in a squat posture.

Muscle fatigue is defined as an exercise-induced reduction in the ability of muscle to generate force or power. Price et al., (2004) provided epidemiological evidence supporting that injury rates tend to be higher at the end of matches and, towards the end of the season, which suggests that muscle fatigue, either acutely (match based) or cumulatively (season based), may be related to knee joint injury. This is hypothesized, as muscle fatigue results in a decrease in muscle force production (Gehring et al., 2008), and therefore knee joint muscles are less able to promote for stability at the knee joint during muscle fatigue. In fact, muscle fatigue is associated with many altered functions of a normal moving knee joint, including decreased knee joint proprioception, reduced pre-activation of the medial and lateral hamstrings and the gastrocnemius muscle, and increased joint laxity (Enoka & Duchateau, 2008, Gehring et al., 2008). Additionally, muscle fatigue is associated
with an increase in anterior tibial translation, which puts direct strain on the ACL (Chappell et al., 2005). In fact, Chappell et al., (2005) hypothesized that fatigued muscles have a decreased capacity to absorb energy as the muscle activation time increases, resulting from an altered neuromuscular function. During muscle fatigue, inadequate shock attenuation capability of the lower extremity joints occurs. There is a decreased capacity of muscles to absorb energy associated with landing impact which results in decreased muscle support about the knee joint (Yeow et al., 2011). Furthermore, muscle fatigue has been shown to cause increased valgus angles about the knee joint, increased knee joint extension angles at landing from a jump and an increase in internal rotation, all of which increase anterior translation of the tibia and therefore strain on the ACL (Chappell et al., 2005).

Although the quadriceps and hamstrings are the main force generators about the knee joint during a maximal vertical jump, and based on their force potential they play a vital role in the mechanical stability associated with the knee throughout the entire task from takeoff to landing. Specifically, during the landing phase, the quadriceps and hamstrings play a critical role in promoting stability at the knee joint as the force they produce protects the passive structures of the knee by reducing anterior translation (Gehring et al., 2008). Therefore, fatigue of these muscles causes considerable constraints to normal knee joint function (Enoka & Duchateau, 2008). The ability of the muscles to create tension in response to landing impact is important and factors such as fatigue that reduce this capacity may as well lead to disruption of normal knee joint function (McLean et al., 2007). Maximal vertical jumps produce fatigue about the quadriceps, hamstrings, and
shank muscles during the takeoff phase, as well as during the landing phase, which make this exercise optimal to study in a progressive manner as each phase of the jump progressively acts in producing fatigue.

To date, research on the effects of muscle fatigue on knee joint function has focused on landing performance in a pre and post-test fatigue intervention. Specifically, to induce muscle fatigue a multi-task protocol was employed, examining landing performance before and after a fatiguing intervention. Fatigu ing interventions ranged in the literature with many different methodologies being utilized to induce muscle fatigue, however they are similar in the sense that the fatiguing task did not include a jump-landing task. The validity of the pre-post-test method, and how accurately it reflects the true effects of fatigue on knee joint mechanics remains unknown. Adapting a progressive same task fatigue protocol to address the current gap in literature is hypothesized to more accurately represent muscle fatigue related changes to jump landing.

Based on work by Ferris, Liang and Farley (1999), humans can rapidly adjust their leg and knee joint stiffness between environmental conditions. With the ability humans maintain in rapidly adjusting joint stiffness between tasks, an important limitation in current literature is revealed. It cannot be assumed that in a pre-post-test protocol that muscle groups used to maintain task performance in a same-task fatiguing protocol have been adequately fatigued. Additionally, when a multitask protocol is utilized, it cannot be assumed that fatigue has occurred in the muscles of interest through the post-test protocol. However, due to the lack of knowledge about the same-task fatigue protocols, we currently cannot validate pre-post-test fatigue
related changes in jump landing.

Muscle force generation involves electrical, chemical, and mechanical systems interacting in concert. The relative contributions of each process are task and time dependent and therefore, should be tested utilizing a protocol, which the participant is progressively fatigued while completing a single task from start to finish (Ferris, Liang & Farley, 1999). Based on the review of the pertinent literature it is evident that our knowledge related to same-task fatiguing mechanisms is limited. Therefore, much is to be learned from examining knee joint mechanical behavior following progressive muscle fatigue caused by a same-task protocol, especially given the high incidences of reported non-contact knee joint injuries.

1.2 Statement of the Purpose

This work will investigate the effects of progressive muscle fatigue on the mechanical behavior of the knee joint throughout maximal double leg vertical jumping. Specifically, we will examine the knee joint of participants performing vertical jumps at their maximum effort continuously at 5-second intervals until fatigue, where the participant perceives they can no longer continue due to muscle fatigue in the lower extremity, or when the trial is ended by the researcher due to perceived injury risk of the participant during the jump landing. This mechanical behavior response will specifically examine if the effects of muscle fatigue are related to sex differences as well as, fatigue level. This work will help fill the current gap in understanding knee joint related injuries associated with progressive muscle fatigue. The knowledge gained can be used in training of recreational athletes to adapt their training protocol. Alternatively, in sport-specific training knowledge
gained from this research can be used to accommodate and strengthen knee joint muscles involved in altering of knee joint mechanics to avoid knee joint injury while exercising in a fatiguing manner.

1.3 Hypotheses:

1) **Females will have greater decreases in overall knee joint flexion angle, and greater increases in knee joint valgus and internal rotation angles, during the landing phase, as fatigue progresses. Fatigue-related changes in knee joint flexion angle will be seen in both sexes, however, females will experience larger changes with time, compared to males. This will be evaluated with a 2x6 mixed ANOVA showing a statistically significant main effect (p<0.05).**

   Current literature, regarding knee joint injury epidemiology (Gage et al. 2012), indicates that females have a higher risk of non-contact knee joint injury than males. Borotikar et al., (2008), in a multi-task fatiguing study, found that females land from a vertical jump with a more extended knee joint, a larger valgus knee joint angle, and a larger angle of internal rotation in a fatigued state, which are all stressors on the ACL and capable of inducing a non-contact knee joint injury.

2) **As fatigue progresses, there will be a significant increase in normalized sEMG for vastus lateralis (VL), vastus medialis (VM), biceps femoris (BF), semitendinosus (ST), gastrocnemius lateral (GL), gastrocnemius medial (GM) and Gracilis (GR), yet there will be a decrease in rectus femoris (RF). Fatigue-related changes in muscle activation will be seen in both sexes (p<0.05),**
however, post hoc tests will reveal that females experienced larger changes with time, compared to males.

The VL (flexor) and the semimembranosus (extensor) have the greatest maximal capacity to produce a moment about the knee joint (Delp et al., 1990). Derouin and Potvin (2005) determined the individual contributions of the 13 muscles crossing the knee joint to knee joint rotational stiffness, about the valgus-varus and flexion-extension axis. They found that the VL and SM also have the greatest potential to stabilize the knee joint in squat postures. Similar results are expected in this study, with the activation of the VL to be the greatest (SM, due to spatial restrictions, was not measured), and maintain their maximal contribution to vertical jumping throughout a progressive fatiguing protocol. It is hypothesized that, as fatigue progresses, activation of the VL and VM will increase (p<0.05). Alternatively, based on work by Sutherland, La Delfa & Potvin (2013), it is hypothesized that the activation of the RF will significantly decrease as fatigue progresses. Ultimately, this will result in an increase in VL and VM activation during jump landing, to maintain the quadriceps role in promoting stability about the knee joint.

3) Fatigue, from repetitive jumping, will result in changes in technique resulting in a decrease in effective duration. It is hypothesized that there will be a decrease in effective duration for both sexes as time progresses. However, there will be an interaction effect between sex and time, ultimately revealing that, as fatigue progresses, female participants decrease significantly greater (p<0.05) then males.
Landing related injuries are common in sports where jumping is a required element. In these sports, the energy from the action of landing is primarily absorbed in the lower extremity joints such as the ankle, knee and hip (Yeow et al., 2011). Specifically, during jump landing, high forces that result from a decreased exertion time, due to a decrease in knee flexion angle, can lead to knee joint injuries (Yeow et al., 2011). Yeow et al., (2011) showed that the first peak of the vertical component of ground reaction force ranged from 1 to 2 kN pre-test, whereas the second peak ranged from 1 to 6.5 kN in landings which occurred after vertical jumps post-test in a fatigued state. When landing forces are elevated above injurious levels, and the knee joint becomes more compromised due to a decrease in knee joint flexion angle experienced during fatigue, the risk of injury elevates. Previous vertical jump landing research, which induced fatigue in a non-progressive manner, demonstrated that maximum vertical ground reaction force values can reach levels as high as 14.4 times body weight for single-leg landings. While fatigued, an increased impact force is experienced by both sexes, and females exhibit a significantly smaller knee joint flexion angle at landing (Chappell et al., 2002). Due to these factors, a decrease in duration of exertion can be expected, which ultimately leads to a decreased effective duration.
Chapter 2
Literature Review

2.1 Knee Anatomy:

In order to understand the effect that fatigue produces through a progressive maximal jumping task on the knee, it is important to first understand and describe the underlying structures that comprise the anatomy of the knee.

2.1.1 Overview:

The knee joint is comprised of 3 bones, specifically, the distal end of the femur, proximal portion of the tibia and the articulating surface on the posterior patella. The knee joint has 5 primary ligaments which function in promoting support of normal knee motion, the anterior cruciate, posterior cruciate, lateral collateral, medial collateral, and the patellar ligament. Also, the 2 menisci of the knee joint, the medial and lateral menisci, function to absorb impact in the knee joint as well as to deepen the knee joint and therefore function to promote and increase stability. Finally, the knee joint contains many accessory structures including the medial and lateral patellar retinacula, the infrapatellar fat pad, the oblique and arcuate popliteal ligaments, the transverse ligaments of the menisci, and finally the bursae. These structures will be detailed below.

2.1.2 Knee Joint:

The distal femur and proximal tibia form the tibiofemoral joint (the weight bearing knee joint), while the femur and patella form the femoropatellar joint (also may be referred to as patellafemoral joint). The tibiofemoral joint links the segments of the thigh to the lower leg. The thigh is the area comprised of the femur and all muscles,
and structure between the hip joint and distal femur. The lower leg is comprised of the tibia, fibula, and all muscles and structures between the proximal tibia to the distal tibia (ankle joint) (Seeley et al., 2008). Both joints are found within a common joint capsule and have communicating articular cavities (Schuenke et al., 2006).

The knee joint is a modified hinge joint that allows flexion/extension, minor valgus/varus motion and slight internal/external rotation, yet the structures which support the knee (muscles, ligaments, bursa) continuously act to promote stability and controls under a great range of loading conditions (Goldblatt and Richmond, 2003).

2.1.3 Femur:

The femur is the longest, strongest bone in the human body, and contains articulating surfaces at the distal end for the weight-bearing joint of the knee (Seeley et al., 2008). The articular surfaces of the distal medial and lateral condyles are pulley shaped to accommodate the sesamoid shape patella and, are convex in both the sagittal and frontal planes (Seeley et al., 2008). The distal end of the femur the tibiofemoral joint is formed with the proximal end of the tibia of the lower leg. The distal end of the body of the femur widens significantly above the knee joint to form the rounded, smooth medial and lateral condyles that contain the articulating surfaces for the tibia. The medial and lateral condyles of the femur meet with the reciprocally formed medial and, lateral condyles of the tibia to form the articular surfaces of the tibiofemoral joint. Between the condyles on the distal femur lies the intercondylar notch, a depression which provides space for the anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL), as well as the accessory
structures of the knee joint, including the medial and lateral patellar retinacula, the infrapatellar fat pad, the oblique and arcuate popliteal ligaments, the transverse ligaments of the menisci, and finally the bursae (fluid filled sac surrounding the knee), used in stabilizing the knee joint along the anterior and posterior axis (Schuenke et al., 2006).

Figure 1: Anterior (A) and posterior (B) aspects of the articulating surface of the patella, tibia and the femur. Also, the articulating surface of the femur to tibia is seen, with many accessory structures labeled such as ligaments, and menisci (Morton, 2011).

2.1.4 Tibia:

The tibia, the larger and stronger of the two lower leg bones (Tortora & Nielsen 2013), is located on the anteromedial aspect of the lower leg. Located distally to the femur (Figure 1), the proximal end forms the tibiofemoral joint with the distal end of
the femur on the medial and lateral femoral condyles. It forms the load-bearing aspect of the knee joint with the femur, as it is reciprocally curved to the femoral condyles to allow for stability and articulation. Between the condyles is the intercondylar eminence, including the tibial spine, which provides attachment points for the menisci and anterior and posterior cruciate ligaments (ACL and PCL). The intercondylar eminence is oriented in an anterior-posterior direction and fits into the intercondylar notch between the femoral condyles. The planed shape of the intercondylar eminence acts as a pivot joint, allowing axial rotation of the tibia through a vertical axis that passes through the medial intercondylar spine (Tortora and Nielsen, 2013). The anterior tibia is also the attachment site of the patellar ligament, as well as important knee extensor muscles on the tibial tuberosity (Tortora and Nielsen, 2013).

2.1.5 Patella:

The patella, commonly referred to as the kneecap, is a triangular shaped sesamoid bone completely embedded within the quadriceps femoris tendon, and is the third bone of the knee joint. The apex of the patella is oriented inferiorly, with the two articulating facets oriented medially and laterally (Figure 2). On the posterior surface of the patella the medial and lateral articular facets are positioned respectively to articulate with the medial and lateral condyles of the femur.
**Figure 2:** The patella in an anterior and posterior view, highlighting the shape of the patella anteriorly, and the articulating surface of the patella posteriorly (Adapted from Tortora and Nielsen, 2013)

### 2.1.6 Ligaments:

Ligaments are defined as a short band of tough, flexible, fibrous regular connective tissue organized into bundles of parallel fibers and function to connect bone to bone or, bone to cartilage (Seeley et al., 2008). Generally, ligaments function to promote mechanical stability for a joint, as well as to support the joint along with the accessory structures and muscles. Ligaments are obligated to function in a state of tension, with its functional position being considered as the position in which the vast majority of its fibers are taut (Fuss, 1989). This state of ligament tension, along with the cross-sectional area and length, is dependent on the joints normal full range of motion during non-injurious movements. Specifically, whether or not the distance between the origin and insertion of the ligament fibers remains constant or, variable as the knee joint moves throughout its normal range of motion (Fuss, 1989). The variability of the distance between origin and insertion is dependent both on the bone attachment point on the ligaments and, design of the tibiofemoral articular surfaces. At lower applied loads the role of the ligaments are to maintain structural
integrity while the muscles guide the joints through their normal range of motion. Additionally, at higher loads, ligaments provide support to limit joint motion to help prevent the segment reaching injury thresholds. Ligaments also assist the accessory structures associated with the joint in the protection of soft tissues during both normal and pathological knee motions (Smith et al., 1993). Of the knee joint there are 5 primary ligaments that will be discussed, including 2 cruciate ligaments, 2 collateral ligaments, and finally the patellar ligament (Figure 3).

Figure 3: Primary ligaments of the knee, as well as the shape of the menisci (Behnke, 2012).

These ligaments, with the assistance of two menisci (articular dense irregular cartilage found on the proximal tibial surface, on the medial and lateral condyles) and joint capsule, located laterally to the menisci, are responsible for promoting
passive joint mechanical stability during normal ranges of motion. The ACL provides 86% of total resistive forces towards anterior translation, and the PCL provides up to 95% of restraining force to posterior translation (Butler et al., 1980). There are several other ligaments and accessory features found within the knee joint which serve to provide the remaining secondary restraint to mechanical stability of the knee joint during motion, typically less than 3% each (Butler et al., 1980), and will not be detailed as extensively. These include the medial and lateral patellar retinacula, the infrapatellar fat pad, the oblique and arcuate popliteal ligaments, the transverse ligaments of the menisci, and finally the bursae (fluid filled sac surrounding the knee).

2.1.7 Cruciate Ligaments:
The cruciate, cross-shaped, ligaments are located intracapsularly and extra-synovially, and are structures that traverse the knee joint, with attachments on the tibia and the femur (Tortora and Grabowski, 1996). The cruciate ligaments are comprised of two ligaments that cross inside the knee forming the anterior cruciate and posterior cruciate ligament. Isometric fibers of both the posterior and anterior cruciate ligaments act together to provide knee joint stability as well as, guiding bundles that control the rolling and gliding motion of the tibiofemoral joint (Fuss, 1989). The anterior cruciate ligament (ACL) extends posteriorly and laterally off the tibia from a point anterior to the intercondylar area to the posterior part of the medial surface of the lateral condyle of the femur. The ACL’s primary function is to limit hyperextension of the knee joint, which is not a normal function of knee joint motion, and prevents the anterior sliding of the tibia on the femur (Tortora and Nielsen,
The normal function of the knee joint is critically dependent on the anatomy of the ACL, due specifically to the position, shape, and dimensions of its areas of attachment (Fuss, 1989). The ACL consists of fibers of varying lengths attaching the tibia and femur, not as a singular unit, but as a collection of individual fascicles that are spread out over a broad area (Fuss, 1989). The various fascicles function to create support of the knee joint through a series of different, normal knee joint functions. The anteromedial fibers form the shortest band of the ACL and are tense in flexion. The remaining bulk of fibers originate from the distal femoral attachment, and insert posterolaterally on the tibia. This posterolateral band is tight in extension and relaxed in flexion (Goldblatt and Richmond, 2003). The orientation of the ACL becomes nearly horizontal on the transverse plane with flexion, and the anteromedial band becomes taut almost immediately after flexion begins. The greater the degree of knee joint flexion, the more horizontal the ACL fibers become. The more horizontal the orientation of the ACL, as seen with greater degrees of flexion, enables the ligament to function as a primary restraint to anterior tibial translation (Goldblatt and Richmond, 2003).

Alternatively, the posterior cruciate ligament (PCL) extends anteriorly, and medially from the tibia and lateral meniscus from a depression on the posterior intercondylar area to the anterior part of the lateral surface of the medial condyle of the femur (Tortora and Nielsen, 2013). The primary function of this ligament is to keep the tibia from posteriorly translating out from under the femur, as well as, acting as a secondary stabilizer to varus-valgus movement and, external rotation (Girgis et al., 1975). The vascular supply to the PCL is from the middle genicular...
artery, which arises from the popliteal artery behind the popliteal surface of the femur. This artery supplies blood to both cruciates, synovial membrane, posterior capsule, and the epiphyses of the tibia and the femur (Goldblatt and Richmond, 2003).

2.1.8 Patellar Ligament:
The patellofemoral joint is found between the anterior surfaces of the distal femoral condyles and the corresponding articular surfaces on the posterior aspect of the patella. The patellar ligament, sometimes referred to as a tendon, is a connective tissue structure that connects the patella to the tibial tuberosity. The patella translates superiorly when the knee joint is in extension, and inferiorly in knee joint flexion. The position of the patella is maintained through its connection to the quadriceps tendon above and the tibial tuberosity below, through its embedding in the patellar ligament.

2.1.9 Collateral Ligaments:
The medial and lateral collateral ligaments (MCL, LCL) support the knee joint in the frontal plane. The primary responsibility of the collateral ligaments is to provide medial and lateral translational, and rotational stability about the knee joint.

The MCL, also referred to as the tibial collateral ligament, is a broad flat ligament on the medial surface of the knee joint, extending from the medial condyle of the distal femur to the medial condyle of the tibia. Tendons of the sartorius, gracilis, and semitendinosus muscles, all of which strengthen the medial aspect of the joint, cross superficially to the ligament. The medial collateral ligament is firmly attached to the medial meniscus (Tortora and Nielsen, 2013).
The LCL, sometimes referred to as the fibular collateral ligament, is a strong, rounded ligament located on the lateral surface of the knee joint extending from the lateral condyle of the distal femur to the lateral head of the proximal fibula. Its function is to provide stability and strength to the lateral side of the knee joint. The tendon of the bicep femoris muscle covers the LCL superficially and the tendon of the popliteal muscle runs deep to it (Tortora and Nielsen, 2013).

2.1.10 Accessory Structures:
The medial and lateral menisci, C- and O-shaped discs of dense irregular fibrocartilage respectively, provide shock absorption and help to improve the fit between the femoral and tibial condyles, by deepening the groove the femur sits in (Goldblatt and Richmond, 2003). These two structures have a large inferior surface and a concave superior surface. The menisci function also to deepen the articulating surface of the knee joint, as well as for stabilization (Steele, 1993).

At the anterior end of the medial meniscus a C-shaped piece of fibrocartilage is attached to the anterior intercondylar fossa of the tibia, anteriorly to the ACL. Its posterior end is attached to the posterior intercondylar fossa of the tibia between the attachments of the posterior cruciate ligament and, lateral meniscus (Tortora and Nielsen, 2013).

The lateral meniscus, a nearly complete “O” shape, shares in the same functions of the medial meniscus, in shock absorption, deepening the articulating surface, and stabilization. (Steele, 1993). Its anterior end is attached anteriorly to the intercondylar eminence of the tibia, and laterally and posteriorly to the ACL. Its
posterior end is attached posteriorly to the intercondylar eminence of the tibia, and anteriorly to the posterior end of the medial meniscus.

The anterior surfaces of the medial and lateral menisci are connected to each other by the transverse ligament of the knee and, to the margins of the head of the tibia by the coronary ligaments (Tortora and Nielsen, 2013).

2.1.11 Muscles:

Superficially to the deep structures of the knee joint, which is comprised of bones, ligaments and accessory structures, are the muscles of the knee joint. The muscles that cross the knee joint from the thigh to the lower leg are activated in order to produce force, which ultimately leads to movement at the knee joint. Alternatively, these muscles along with the ligaments, menisci, and accessory structures, function synergistically in order to promote stability about the knee joint throughout the joint's normal full range of motion. The muscles that control flexion and extension of the leg at the knee joint originate superiorly to the knee joint and insert inferiorly to it. Note that some of the muscles of the knee are biarticular (meaning that they cross two joints), as they also cross the hip joint that enables movement about that joint.

To understand the role that muscles provide in mechanical joint stability it is necessary to discuss the muscles responsible for motion about the knee joint. The following section has been adapted from three primary sources (Tortora and Nielsen, 2013; Seele et al., 2007, and Schuenke et al., 2006). Additional resources, when required to add detail or highlight supplementary information not provided by the above three sources, will be cited accordingly.
2.1.12 Muscles of the Knee:
The primary muscles involved in flexion and extension of the knee joint that will be discussed include the quadriceps group, located in the anterior compartment of the thigh, and the muscles that make up the hamstring located in the posterior compartment of the thigh. It is also important to note that along with flexion and extension, these muscles function to promote mechanical stability about the knee joint. Motions such as flexion/extension, internal/external rotation and valgus/varus rotations of the knee joint allow the body to perform such important movements as walking, running, kicking, and specifically to this research, maximal vertical jumping.

2.1.13 Anterior Compartment:
Found on the anterior surface of the thigh are the four muscles which make up the quadriceps femoris group. The quadriceps femoris (QF) is a combination of muscles comprised of four distinct muscle groups: rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), and the vastus intermedius (VI). The heads of the QF share a common point of insertion onto the patella, and through the patellar ligament finally inserts onto the tibial tuberosity. Contraction of the quadriceps group produces extension about the knee joint as well as, causes extension of the leg about the hip joint.

The bi-articular sartorius muscle, the longest muscle in the human body, crosses the anterior compartment muscles from the hip, however plays a role in weakly flexing the leg at the knee. The sartorius originates at the anterior superior iliac spine, and inserts on the medial surface of the body of the tibia. As the sartorius is a bi-articular muscle, it has an alternate function of weakly flexing, abducting and
laterally rotating the thigh at the hip joint.

2.1.14 Posterior Compartment Muscles:

The hamstring muscles are located on the posterior aspect of the thigh: semitendinosus (ST), semimembranosus (SM), and biceps femoris (long head (BF-L) and short head (BF-S)). The hamstring muscles originate from the ischial tuberosity, with the exception of the biceps femoris short head which originates on the lateral condylar ridge of the femur. The hamstrings are a bi-articular muscle crossing both the knee and the hip joint and, are primarily responsible for knee joint flexion, knee joint internal and external rotation, as well as, hip extension when the trunk is in a fixed position.

The biceps femoris (BF) is comprised of a short and long head that have a common insertion onto the head of the fibula and, the lateral condyle of the tibia. The biceps femoris short head originates on the lateral lip of the linea aspera, while the biceps femoris long head originates on the ischial tuberosity. The BF short head only acts at the knee, flexing the leg on the thigh. Both BF muscle groups are oriented in a medial to lateral direction. Due to this orientation, the BF also is responsible for lateral rotation of the leg (Figure 4).

The semitendinosus inserts on the medial surface of the superior aspect of the tibia. The orientation of its fibers runs in a slight lateral to medial direction which creates medial rotation about the knee joint when both the leg and thigh are flexed (Figure 4).

The semimembranosus inserts onto the posterior aspect of the medial condyle of the tibia. The orientation of its fibers travels in a lateral to medial
direction, similar to the ST, and also functions to medially rotate the leg about the hip joint (Figure 4).

**Figure 4:** Anterior and posterior compartment muscles of the quadriceps, hamstrings and, of the lower leg, calf muscles (Vella, 2008).

### 2.1.15 Gastrocnemius:

The gastrocnemius is the most superficial of all the posterior compartmental muscles of the lower leg and has a common insertion via the achilles tendon onto the posterior surface of the calcaneus. The gastrocnemius is a bipennate muscle, meaning it has two heads, and is a biarticulate muscle. The gastrocnemius medial
head originates on the popliteal surface of the femur, superior to the medial condyle. The gastrocnemius lateral head attaches proximally onto the lateral aspect of the lateral condyle of the femur. Collectively, both heads function to flex the knee joint and plantar flex the ankle (Figure 4).

2.2 Knee Injury Epidemiology: The knee joint is the most commonly injured joint by adolescent athletes where an estimated 2.5 million sports-related injuries are reported by emergency departments annually in the United States (Gage et al., 2012). Of all of the reported injuries associated with the mechanical structures of the knee joint, ACL injury rates are the most common (Donnelly et al., 2012). Disruption to the ACL has become so common that the seriousness of the injury is often overlooked. This ligament is stretched or torn in about 70 percent of all serious knee joint injuries (Tortora and Nielsen, 2013). A ruptured ACL is considered one of the most severe knee joint injuries an athlete can sustain in sport (Donnelly et al., 2012), and has negatively affected many professional and recreational athletes.

It is approximated that one out of three of all sports injuries resulting in surgery involves damage to the ACL (Powell et al., 1999; Rishiraj et al., 2009), making it the most costly procedure when all surgery and rehabilitation expenses are considered (de Loes et al., 2000; Louw et al., 2008). This has resulted in countless studies examining ACL injury prevention strategies, yet it has been suggested such focus has not been effective in resulting in lower incidences among athletes competing in recreational athletics (Donnelly et al., 2012). These high rates of knee joint injuries have resulted in a massive financial burden as it has been estimated that over $1.5 billion are spent on these injuries in the United States alone.
Along with a financial burden, data has shown that approximately 55% of ACL injured athletes are not capable of returning to the same level of competition two years post-reconstruction, a percent that increases to 70% after 3 years (Donnelly et al., 2012).

Although knee joint injuries pose an economic burden on the injured individuals, the healthcare system is also burdened by the quantity of knee joint injuries, as they are so frequently seen in United States emergency departments (Gage et al., 2012). As the data from Gage et al., (2012) is based off of a survey taken from knee joint injuries presented only to emergency rooms across the United States, this study is hypothesized to underestimate true knee joint injury rates. In reality, the clinical and public health burden of knee joint injuries are likely much higher than reported from emergency room visits, as individuals with knee joint injuries may seek care from a wide variety of clinicians (athletic trainers, primary care physicians, urgent care centers, sports medicine clinics, etc.).

It was estimated that there was over 6.6 million knee joint injuries presented to United States emergency departments from 1999 through 2008, for a rate of 2.29 knee joint injuries per 1,000 in the population (Gage et al., 2012). Certain demographics proved to be more at risk of a knee joint injury then others, specifically those aged 15 to 24 years of age who had the highest injury rate (3.83), while children younger than 5 years had the lowest rate (0.55). Most commonly seen injuries affecting the knee joint were strains and sprains making up 42.1%. The most common mechanism of knee joint injury resulting in an emergency department visit was sports and recreation at 49.3% (Gage et al., 2012). The current literature on knee joint injuries, although extensive, focuses primarily on young athletes,
particularly the female population. Several sex and age group differences were identified by Gage et al., (2012). For example, knee joint injuries were reported to account for 60% of high school sports-related surgeries, and female athletes have been reported to be four to six times more likely to sustain a major knee injury. There are many factors identified in the research that may attribute to females having a higher risk of knee joint injuries, including anatomical body structure, environmental, hormonal, neuromuscular, and biomechanical differences (Gage et al., 2012). de Loes et al., (2000) reported that female athletes have twice as many ACL injuries, and it has been subsequently reported that females require surgery for ligamentous knee joint injuries almost twice as often as males (Fernandez et al., 2007). The significantly higher number of ACL injuries among female (compared to male) athletes has prompted a focus on exercise interventions in an attempt to reduce the risk of injury specifically in the female population, nonetheless towards males as well.

Although there are many factors implicated in ACL injury, ultimately, when forces associated with the loading demand on the ACL becomes greater than its physical capacity to withstand those demands, risk of ligament damage is at its greatest. There are two primary mechanisms of injury that can occur to the knee joint, a contact injury, resulting from an external impact of another individual, or non-contact injury, resulting from no contact with any other object or person except for the ground, such as landing from a jump or pivoting while running. Therefore, it is important to understand both the mechanisms associated with increased loading of the ACL, and its physical capacity in order to best prevent ACL injuries from contact or non-contact loading (Dempsey et al., 2011). Yeow et al., (2011) and Liederbach
et al., (2008) indicates that 70-90% of sports injuries occur in non-contact situations. An epidemiological study by Morgan et al., (2014) showed over 400,000 ACL injuries occur in the United States annually, with approximately 80% of these being non-contact ACL injuries, most frequently occurring during single leg jump landing, or side-stepping sport tasks. Cochrane (2007) labeled the two most common sporting maneuvers observed during non-contact ACL injuries to be generalized as sidestep cutting and landing, following a jump. It is critical in the understanding of non-contact ACL injuries to study these two actions as, they have been shown to produce high valgus and internal rotation moments about the knee joint, coupled with anterior draw (anterior translation). The moments are caused by quadriceps extension, which can drastically increase the force demand on the ACL, particularly in extended knee joint postures (Dempsey et al., 2011) found during fatiguing situations. It appears that extended and internally rotated lower limb postures, with an abducted hip, are associated with increased peak valgus moments (Dempsey et al., 2007), a known mechanism of injury for non-contact ACL injury. This would suggest that these postures have an increased risk of ACL injury.

Functionally, the ACL is the primary restraint to anterior translation of the tibia, as it provides an average of 86% of the total resisting force to anterior drawer (Butler et al., 1980). The intersection of ACL together with the PCL defines the axis of rotation of the knee joint. The multidirectional forces imposed on the complex knee joint during physical activity explain the types of severe knee joint injuries, including ruptures of the ACL or PCL, commonly seen in reported knee joint injuries. As previously stated, prevention strategies are important as a knee joint injury can end an athlete’s competitive career and, can permanently affect physical activity
capabilities among individuals of all ages.

When the knee joint is near full extension in a jump landing, experienced in such sports as volleyball, basketball and gymnastics, the application of externally applied translation forces, coupled with valgus and internal rotation knee joint moments, results in forces on the ACL up to, or above physical capability thresholds. Such thresholds for knee joint demand have been listed as >2160 N by Yeow et al., (2011). Specifically during landing these forces, coupled with an out-of-plane knee joint motion, can lead to articular cartilage damage, ligament ruptures, bone bruises and menisci tears (Yeow et al., 2011). The mechanics of the normal knee joint is to provide support and promote stability at any given exertion, under normal human body mechanics. When these forces are elevated above injurious levels and more-so compromised in out-of-plane (deviating from normal knee-joint motion) knee joint positions, the risk of injury elevates. Previous vertical jumping studies relating to impact forces have shown that maximum vertical ground reaction force values can reach as great as 14.4 times body weight for single-leg landings. Yeow et al., (2011) showed that the first peak of the vertical component of ground reaction force ranged from 1 to 2 kN, whereas the second peak ranged from 1 to 6.5 kN in landings which occurred after a jump similar to a volleyball block jump. As previously reported by Morgan et al., (2014), the injury threshold for the ACL has been reported as >2.16 kN.

Accumulation of excessive and repetitive impact forces may pose a threat to the integrity of the lower extremity. Many of these repetitive impact injuries are commonly associated with the knee joint, and are most commonly reported in running, basketball, tennis, volleyball, skiing, and triathlon (all of which are non-
contact sports, including side-stepping and/or jumping). Additionally, in a sport such as basketball research indicated that 58% of all injuries suffered by female players occurred while landing from a jump. In addition, it has been reported that knee joint pain is the most common problem experienced by runners, which can translate to any sport which involves large quantities of running (Zhang et al., 1998). It is important to note that the hip, knee and ankle joints all contribute to shock absorption via energy dissipation by the muscles, however the primary focus of this literature review is to examine the effect on the knee joint. Decker et al., (2003) found that the knee joint was the primary energy absorber for both sexes during double leg landings. Also, it has been shown that the ankle plantar flexors and, hip extensors were the second largest contributors to energy absorption for the females and males, respectively (Yeow et al., 2011).

2.3 Effects of Fatigue on the Knee during Jumping and Landing:

Enoka & Duchateau (2008) describe muscle fatigue occurring during voluntary movement as an exercise-induced reduction in the ability of muscle to produce force or power. In simple terms, muscle fatigue represents a reduction in the ability to generate force or power (Gandevia, 2001). Muscle fatigue results from an accumulation of actions from the human body performing work and thus, reduces the capacity of healthy individuals to produce their maximal force output (Enoka & Duchateau, 2008) and, is associated with many altered functions of all joints including the motion of the knee including; decreased knee joint proprioception and, increased joint laxity compared to baseline values. Subsequently, muscle fatigue is frequently cited as a risk factor for knee joint injury during landing tasks, in fact,
Gehring et al., (2009) found muscle fatigue directly related to increased incidences of ACL injury. Epidemiological evidence shows that the injury rates tend to be more frequent approaching the end of a physical activity session, which suggests that fatigue could in fact relate to injury. Multiple biomechanical parameters including lower extremity kinematics and kinetics, neuromuscular factors such as muscle co-activation, and also proprioceptive function can be influenced by fatigue and therefore, are believed to lead to injury (Gehring et al., 2009). Gehring et al., (2009), also found that fatigue led to a reduced pre-activation of the medial and lateral hamstrings and, the gastrocnemius muscle. This evidences helps to provide evidence that there is different knee joint control in fatigued conditions, leading to a hypothesis that muscle fatigue may influence non-contact knee joint injuries. Leiderbach et al., (2008), reported that dancers suffered considerably fewer ACL injuries than athletes participating in other sports involving jumping and side-stepping. It is speculated that the postural training associated with dancing may actually serve to protect them against ACL injury even in fatiguing situations. As previously stated, adapted from Gehring et al., (2009), non-contact knee joint injuries most frequently occur near the end of a game or practice, leading researchers to hypothesize an effect of fatigue.

Muscle fibers have been shown to have a decreased capacity to absorb energy when fatigued, and have altered neuromuscular function. Specifically, muscle fatigue is associated with increased anterior tibial translation, directly straining the ACL (Chappell et al., 2005). In addition, increases in valgus, extension and, internal rotation angle (all associated with increases in anterior translation of the tibia, and therefore strain on the ACL), have all been connected with knee joint
injury and when coupled with muscle fatigue, increases the risk of knee joint injury (Chappell et al., 2005). As muscles fatigue, inadequate shock attenuation capability of the lower extremity joints has been shown during impact at landing (Yeow et al., 2011), and this is hypothesized to lead to articular cartilage damage, ligament ruptures, bone bruises and menisci tears.

Fatigue of the knee joint musculature is hypothesized to directly lead to ACL injury risk via the sacrifice of joint mechanics to maintain mechanical demand. These strategies are hypothesized to accommodate mechanical demand during physical activity. However, there remains a lack of knowledge related to the association of muscle fatigue and mechanical behavior of the knee joint during landing impacts. More specifically the effects of progressive fatigue about the knee joint studied on a same-task fatiguing protocol has never been examined. However as documented by Borotikar et al., (2008) in multi-task research, it has been seen that fatigue induced sacrifice of joint mechanics as observed during maximum fatigue, have already become evident before the maximum fatigue level. Progressive fatiguing research will help to provide insight towards the effects that fatigue has on body mechanics, specifically at the knee joint, throughout the protocol, highlighting that participants do not need to be maximally fatigued to experience the negative impacts fatigue produces on body mechanics.

The present review of literature has identified potential limitations within previous research protocols regarding the relation of the results from a pre-post test fatiguing study, to realistic athletic occurrences. Research which has been completed to fatigue participant’s leg muscles in an effort to measure changes in body mechanics, have utilized a pre-post test method. This is believed to be limited
in terms of reliability to real athletic occurrences as fatigue progresses differently, using different muscles, through a single task, then through a multi-task protocol. Although informative, and the basis to this research, we believe that a progressive fatiguing protocol on a single task, vertical jump landing, will best outline the changes experienced through fatigue in a jump landing task. Also due to the lack of research on progressive same-task fatigue, limitations have been seen in progressive monitoring of changes in knee joint mechanics which may lead to non-contact knee joint injury. There is a lack of available research in the literature regarding the effects of progressive fatigue on same-task body mechanics to investigate when the serious consequences of fatigue begin.

Altered knee joint biomechanics, and in particular, deviations from normal movements of the hip and knee, are common when whole-body fatigue is associated with dynamic sport jump landings (Borotikar et al., 2008). In prolonged tasks causing fatigue, such as running and, tasks that involve a rapid deceleration to stop, delayed onset of quadriceps and hamstring muscle activation as well as, early occurrence of maximal knee joint flexion were found (Chappell et al., 2005). Biomechanical changes in body mechanics, as well as, changes in muscle activation decrease shock absorption during impact and, decrease knee joint stabilization during landing (Zhang, et al., 2000) from a vertical jump. Attenuation is decreased at the knee during fatigue, as the knee joint becomes more extended and less flexed during landing, resulting in a harder landing and thus, directly effecting the structures that absorb external forces. Hamstring fatigue, and the resultant increase in muscle activation time, resulted in decreased peak impact knee joint flexion moment (therefore a greater degree of knee joint extension), and increased
internal tibial rotation, both of which are mechanisms of injury for non-contact knee joint injuries (Chappell et al., 2005). Borotikar et al., (2008) also found that hamstring muscle activation decreased after fatigue resulting in a decrease in the hamstring/quadriceps (H/Q) co-activation ratio. A decrease in the H/Q ratio results in an increase in anterior translation, a mechanism of injury for the ACL.

In addition to their function of maintaining the necessary mechanics to land following a vertical jump, the muscles of the thigh and the lower leg aid in the attenuation of the associated impact forces. Muscle fatigue forms a considerable individual limitation to knee joint function (Enoka & Duchateau, 2008), as there is evidence suggesting that fatigue of the knee joint musculature suppresses knee joint muscle reflexes (Wojtys et al., 1996) and, reduces joint stiffness, which has a direct negative effect hypothesized to result in a reduction of knee joint stability. The changes have been attributed to increased electromechanical delay (EMD) (Chan, Lee, Wong, Wong, & Yeung, 2001; Moore, Derouin, Gansneder, & Schultz, 2002) and, reduced stiffness of the muscle-tendon complex (Mair, Seaber, Glisson, & Garret, 1996). A reduction in knee joint stability, hypothesized to be due to fatigue, provides evidence that compensatory mechanisms must be used to counterbalance the loss of the muscle force-generating properties associated with fatigue (Rodacki et al., 2002) to continue an athletic performance. Additionally, Van Ingen Schenau et al., (1995) proposed that changes in muscle activation timing occur to avoid deterioration of the athletic performance of the knee joint when the properties of the musculoskeletal system are changed (Rodacki et al., 2001). From this it can be hypothesized that safe knee joint mechanics are sacrificed in order to gain or, maintain an optimal athletic outcome. Other non-contact studies involving cycling,
running, sprinting, lifting, and continuous hopping exercises in their jumping protocol provide evidence that participants will sacrifice joint mechanics and adapt compensatory mechanism to maintain athletic performance in order to counterbalance the loss of the muscle force generated due to the effects of muscle fatigue, (Borotikar et al., 2008, Yeow et al., 2011, Chappell et al., 2005, Donnelly et al., 2012).

As noted, muscle fatigue may change the strategies used during landing following a jumping task to accommodate the associated impact forces. Borotikar et al., (2008) observed a landing strategy that consisted of an extended knee joint angle when subjects were fatigued, which has been believed to prevent the excessive lower limb collapse, and thus injury that may have occurred. In support of this finding, Chappell et al., (2005) found that lower extremity fatigue is associated with increased peak proximal tibial anterior shear force and, a decreased knee joint flexion angle. Based on such findings, it can be hypothesized that fatigue alters the mechanics associated with the knee joint such that the mechanisms of injury related to non-contact knee joint injuries are amplified. With such findings, it is possible to hypothesize that a decline in performance mechanics after fatigue may be the result of a change in coordination (changing the neural input), a change in the functional capacity of the muscles to produce force or, the combination of these two factors (Rodacki et al., 2001).

Fatigue reduces co-contraction of the lateral hamstrings in the critical phase of touch down which is thought to be responsible for overall knee joint stability (Gehring & Gollhofer, 2009). Alternatively, Chappell et al., (2005) found that the
effect of fatigue on the peak proximal tibial anterior shear force during landing of a jump, resulted in a mean increase of up to 96% in knee joint valgus angle. Also recorded from Chappell et al., (2005), in a series of stop-jump tasks at fatigue, measurements of 0.29 times the body weight at landing, compared to the 0.24 recorded in a rested state. This result represents an average of 21% increase in the peak proximal tibial anterior translational forces at landing when the participant has become fatigued. Rodacki et al., (2001), in an examination of fatigue on vertical jump coordination, found a pronounced decrease in the maximal countermovement jump performance that occurred by fatiguing the knee joint extensor muscles. This suggests that fatiguing of the extensor group possesses greater importance than its antagonist counterpart. Additionally, Rodacki et al., (2001) estimated that the knee joint flexor muscle group exerts a moment of force that accounts for 11% to 17% of the resultant joint moment, which is substantially less than that generated by the knee joint extensors. Fatiguing the knee joint extensor muscles would result in participants having to adjust several kinematic and kinetic variables of the jumping movement, which would include a reduced joint angular displacement of the knee joint, a decreased knee and hip peak joint angular velocity, and an increased knee joint stiffness (Rodacki et al., 2001). Borotikar et al., (2008) found that in a fatigued state, both internal rotation and abduction motions about the knee increased substantially, and these postures are known to increase loading on the ACL. Landing with internal rotation and knee joint abduction (knee valgus angles) likely promotes less than optimal bi-articular quadriceps and hamstring lengths (Delp et al., 1990), compromising their ability to successfully oppose external knee joint abduction loads (Borotikar et al., 2008).
There is insufficient evidence to establish the precise link between muscle fatigue and dysfunction of the knee joint during landing tasks. However, exercise at or near exhaustion, for example, causes increased knee joint laxity (Wojtys et al., 1996), possibly compromising ligament mechanoreceptor feedback and muscle strength. A key characteristic involved in the effects of fatigue is the location and magnitude of observed fatigue effects and, the dependency of the muscle group on the characteristics of the task, known as task dependency (Enoka, 1995). The task dependent progression, which this research aims to focus on a single task, of fatigue is more complex than simple failure of the processes responsible for the mechanical output of skeletal muscles.

Observations have been used to link fatigue to increased risk of knee joint injury (Hughes & Watkins, 2006; Yu et al., 2002) and to assist in the knowledge of associated mechanisms of musculoskeletal injury (Kumar, 2001). Therefore, muscle fatigue may be considered an important individual constraint to knee joint function during jump landing as well as to any other actions in which the knee joint becomes fatigued in recreational athletics.

2.4 Protocols:

To date, the majority of studies examining knee joint function during landing from a maximal jump have assessed performance in pre-post test fatiguing protocols. Specifically, in a pre-post test fatigue protocol, participants complete a supplementary task to induce fatigue for the muscle group being studied between testing sessions. With this method, participants are not progressed towards fatigue from the task being studied, rather have an alternative method of inducing fatigue,
such as using a leg-press machine following the initial pre-test protocol which is completed from rest and, preceding the post-test task. Following the protocol used to induce fatigue, participants resume the testing task to examine the results of fatigue on the initial rested pre-test task. The reliability of this testing method with relation to the method in which fatigue is induced, remains uncertain as fatigue is not induced by the task being examined. The following section shows there is a gap in research concerning the effects of same-task progressive fatigue and its effects on jump landing mechanics.

### 2.4.1 Progressive fatigue protocol vs. before after testing

Two different protocols of fatiguing will be discussed in this section: progressive fatigue and pre-post test fatigue testing. Progressive fatigue requires participants begin the protocol fully rested and then complete an assigned activity in set time intervals that they must maintain until they reach a measureable percent level of fatigue. In contrast, the pre-post test fatigue testing, involves recording the trials of the participant in a rested state, followed by fatigue causing bouts of non-task-specific exercise of the muscle group of interest (such as exercises on standard exercise equipment). Finally, the post-test is completed using the same task as the pre-test protocol, which differs from the task used to fatigue the muscles. Such models, however, do not necessarily reflect how fatigue progresses and implicates during realistic sports participation, suggesting a more general induction strategy is warranted. The method used to induce fatigue in the pre-post test models most often are based on repetitive loading of targeted muscle groups which precipitate substantial movement control similar to that of the testing task (Miura et al., 2004;
Wojtys et al., 1996). However, progressive fatigue models appear to more accurately simulate sport-relevant movement tasks and, may induce both local and central fatigue effects along the entire proprioception pathway (Borotikar et al., 2008).

A modified version of a progressive fatigue protocol was used by Borotikar et al., (2008), in which each subject was required to perform approximately 30 pre-fatigue landing trials. This was followed by a series of jump-landings while being simultaneously exposed to a general progressive fatigue protocol, which included continuous sets of five double leg squats between jump trials. Borotikar et al., (2008) chose to induce fatigue through repetitive squatting tasks, since they hypothesized that this represents a closed kinetic chain exercise involving hip, knee and ankle agonistic and antagonistic musculature, which is equally prominent in the control and coordination of jump landing tasks. From this, Borotikar et al., (2008) quantified and compared lower limb joint kinematics in parallel with the progression towards maximum fatigue (Madigan and Pidcoe, 2003), finding that, overall, fatigue negatively affected males and females at the knee joint due to changes in body mechanics. Although Borotikar et al., (2008) used this progressive multi-task fatigue approach, the protocol did not utilize task specific tasks to induce the fatigue and thus their results may not accurately represent a true sport-relevant situation.

It must be noted that there are concerns in the literature towards adapting a progressive fatiguing protocol. Fatigue is extremely difficult to quantify, and therefore compare fatigue-induced strength and/or power adaptations between participants based on level of fatigue (McLean et al., 2007). To address this issue, recent studies (Chappell et al., 2005) defined maximal fatigue levels based on
volitional exhaustion, the point that participants can no longer perform a specific component of the fatiguing tasks. It is assumed in this case participants indeed have progressed through to a realistic exhaustive end point, and that muscle groups experiencing fatigue are actually utilized within the comparative performance task (Borotikar et al., 2008). This approach may be limited as it fails to focus enough attention on the fatigue levels of the lower limb muscles which, is theorized to directly affect the mechanics of the knee joint. It is hypothesized that examining fatigue as a progressive decrease in jump height through continuous maximal vertical jumps, the muscles being studied around the knee joint will be adequately fatigued for this task-specific research. This is hypothesized to represent a more accurate representation of the effects of fatigue on mechanics at the knee joint during landing from a vertical jump.

The majority of current research investigating relationships between fatigue and joint mechanics have adopted a pre-post test fatigue model, with resulting injury hypotheses based on disparities between pre and post fatigue measures (Chappell et al., 2005; McLean et al., 2007; Nyland et al., 1997, Wojtys et al., 1996). However McLean et al., (2007), suggests that a rapid recovery from fatigue on the muscles being examined may present within such a design, ultimately compromising the outcome and conclusion efficacies. Moreover, McLean et al., (2007) is stating that since a multi-task approach was used to stimulate fatigue about the muscles of interest, fatigue recovery may begin to occur during the post-test protocol, as fatigue was induced by different means (a different task, which does not utilize the exact same muscle recruitment as the testing task), and therefore the effects of fatigue may not be truly representative of the results that fatigue has on body mechanics. It
must be noted that pre-post test fatigue protocols where fatigue is induced with either, very functional exercise protocols or, in strictly controlled (isometric) open-chain conditions, the potential issue of generalizing the fatigue results is not addressed as the fatigue remains induced from a different task then that being studied. For example, functional fatigue protocols using running and jumping have been shown to lack the ability to control the state of fatigue, therefore compromising the results of the study (Chappell et al., 2005). On the other hand, the functional relevancy of open-kinetic-chain fatigue protocols (that focus on fatiguing specific muscle groups) are not typical for running and cutting sports with a high risk of ACL injuries and therefore, knowledge from those works may not aid in understanding ACL injuries (Fagenbaum and Darling, 2003). Gehring et al., (2009), in an attempt to understand knee joint mechanics between sexes during jump landing, used a sub-maximal fatigue protocol in a leg press weight machine, selected due to its correlation of closed-kinetic-chain knee joint flexion and extension movements with other functional fatigue protocols and, because it was possible to define a clear break-off criterion. The subjects were instructed to flex and extend their knee joints (90° to full extension) in the leg-press with 50% of their maximum load (1-repetition maximum), these exercises were continued until the subjects could no longer perform the selected load. Chappell et al., (2005) also utilized a pre-post fatigue protocol where, each participant performed 5 trials for each stop-jump task prior to experiencing fatigue and then immediately after fatigue was achieved. To induce fatigue a customized fatigue exercise was used which consisted of unlimited repetitions of 5 consecutive vertical jumps followed by a 30-m sprint. The vertical jumps required a squat start position followed by a jump where the arms and hands
were extended to reach a target height of 115% of the participant’s vertical reach. The fatigue exercise was continued until the subject reached a state of volitional exhaustion, a point at which a participant feels they can no longer continue and voluntarily terminates the effort, followed by the post fatigue test. It was hypothesized that volitional exhaustion has too many confounding variables making it difficult to determine if the muscles of interest to the research are truly fatigued or if, for example, the participants in the Chappell et al., (2005) work could no longer complete a 30 meter sprint due to cardiovascular capacity, rather than terminating the effort due to maximal fatigue about the muscles of the leg. There is an identifiable gap of knowledge in the difference between progressive same-task fatigue, and what is commonly studied in the literature in pre-post test fatigue measurements, specifically related to real world fatiguing situations for athletes.

2.5 Sex Differences:

ACL injuries are much more common in females than males, perhaps by a factor of as many as 3 to 6 times that as reported by males (Tortora and Nielsen, 2013). Although the root cause of this sex difference remains unclear, there is speculation that the following factors play an important role: (1) less space between the femoral condyles in females thus, limiting space for the ACL movement; (2) the female pelvis is wider, creating a greater angle between the femur and tibia (also known as Q angle); (3) female hormones allow for greater flexibility of ligaments, muscles, and tendons; (4) females have generally shown to have less muscle strength causing an increased dependence on the passive forces produced by the ACL to maintain knee mechanical stability (Tortora and Nielsen, 2013; Laughlin et al., 2001; Hewett et al.,
In fact, Arendt et al., (1995) and Gomez et al., (1996) have found female athletes are more likely than male athletes to sustain ACL injury during non-contact situations such as jump landings. In addition, Borotikar et al., (2008), determined that females land from a vertical jump with a more extended knee joint, a larger valgus knee joint angle, and a larger angle of internal rotation, which are all stressors on the ACL capable of inducing a non-contact knee joint injury.

Knee joint injury research concerning sex is based on distinct differences in injury rates and, how fatigue affects body mechanics associated with non-contact knee joint injuries between males and females (Gage et al., 2012). Although it is well established that the incidence of ACL injuries (contact and non-contact) in females is substantially greater than males, the underlying injury mechanisms remain unclear, especially when individuals are fatigued (Gehring et al., 2009; Chappell et al., 2005). Presently, the association between fatigue and lower extremity injury, including ACL ruptures, remains anecdotal, and the relationship between fatigue and ACL loading in high-risk athletic tasks has not been fully explored. There is speculation that there is an association with ACL injury and stop-jump tasks, and biomechanical analysis of these tasks by Chappell et al., (2005) have shown an increased peak proximal tibial anterior shear forces for females compared to males during the landing phase. In addition, Chappell et al., (2002) showed that females experienced greater abduction angles and, higher abduction moments during landing and cutting movements (Chappell et al., 2002). Such biomechanical contrasts in sex are hypothesized to be caused based on differences in neuromuscular activation strategies, specifically, the imbalances in H/Q activation ratio is thought to be a risk factor for ACL injuries. Several studies have determined
that females show H/Q deficits implying a reduced active joint control strategy compared to males (Hewett et al., 2005). An H/Q ratio below 0.6 is a risk factor and predictor of ACL injury, and females’ lower H/Q angle is theorized to make them more susceptible to non-contact ACL injury (Hewett et al., 2005).

In addition to the aforementioned injury mechanisms, research shows that female participants, when landing from a jump, have significantly greater knee extension moments at the peak proximal tibial anterior shear force than their male counterparts (Chappell et al., 2002). Chappell et al., (2005) showed females had a significantly larger valgus angle at landing and, greater difference in the knee joint valgus-varus moment at the instant in time where peak proximal tibial anterior shear force occurred when landing from a vertical jump. In the same study completed by Chappell et al., (2005) it was reported that, once fatigued, males produced knee joint varus moments at jump landing (which, as stated by Chappell (2005) is more-so protective towards non-contact knee joint injuries). Additionally, Chappell et al., (2005) reported that female’s created significantly smaller knee joint flexion angles during landing at the peak proximal tibial anterior shear force than did males (with knee joint flexion, the ACL is at a lower risk of injury as the ligament is not being strained). It is hypothesized that since fatigue effects sex differently, the increased peak anterior shear force on the proximal tibia can be attributed to females experiencing the effects of muscle fatigue more than males. As per the aforementioned research by Chappell et al., (2005), outcomes in both sexes resulted in smaller knee joint flexion angles in the post fatigue tests, but differences in knee joint valgus-varus moments were observed for males and females, showing that females are at a greater risk of knee joint injury while experiencing muscle
fatigue.

2.6 Joint Rotational Stiffness and Effects of Muscle Mechanics in Fatigue:

Literature shows that under fatigued efforts there is a mechanical disruption of the knee, ultimately resulting in joint injury. Although the actual quantification of mechanical integrity/stability of the knee joint will not be measured in this study, a measure of the associated muscle activation will be. It is important to briefly discuss joint stability, given the fact that muscle fatigue has been shown to negatively affect the mechanical integrity, which is required to prevent structural disruption.

The knee joint, a complex modified hinge joint, is considered the lower extremity joint with the least dependency on the shape of its articulating surfaces to maintain mechanical stability (Derouin & Potvin 2005). Mechanical stability is defined as the ability of a loaded structure to maintain static equilibrium even at small fluctuations around the equilibrium position (Bergmark, 1988). To promote for stability, the large muscle groups of the anterior and posterior compartment of the thigh that also possess high geometric stability (related to having large functional moment arms and short fiber lengths), are the greatest contributors in maintaining knee joint stability through normal human range of motion (Derouin & Potvin 2005).

Potvin and Brown (2005) developed an equation to calculate individual muscle contributions to joint stability based on muscle force, muscle stiffness, and origin/insertion coordinates (Figure 5). Specifically, Potvin & Brown (2005) demonstrated that the stability of a single degree of freedom system was equivalent to its rotational stiffness. Brown & Potvin (2007) continued in this research, showing
that the stiffening capability of muscles provides a first look into their stabilizing potential (Brown & Potvin, 2007). Therefore, if the specific maximum stiffening potential of a muscle is understood, determining which part of the system is most vulnerable to a failure is possible, which can help understand why a mechanical injury to a joint occurs. Once individual muscle forces are estimated about the knee joint for each condition of research, their contributions to tri-axial knee JRS can then be calculated with the following equation (Potvin & Brown, 2005; Cashaback & Potvin 2012):

$$JRS_z = F \left[ A_x B_x + A_y B_y - \frac{r_z^2}{l} + \frac{q r_z^2}{L} \right]$$

**Figure 5**: Equation developed by Potvin and Brown (2005) to calculate Joint Rotational Stiffness

Total JRS (JRST) is calculated by summing individual muscle contribution to JRS. Extensor and flexor contributions to JRS (JRS\textsubscript{E} and JRS\textsubscript{F}) are calculated by summing the muscle contributions of either all of the flexors or, all of the extensors. Additionally, each muscle’s relative contribution to JRS can be determined for each condition by taking an individual muscle contribution and dividing it by the JRST. Since individual muscle force and JRS increase proportionally with EMG activity (Potvin & Brown, 2005), it should be believed that JRST also increased linearly.

A muscle’s orientation about an axis of a joint determines its ‘geometric stiffness’ which, is influenced by moment arm and muscle length; with the former
having the greatest influence (Brown & Potvin, 2007). The VL and the SM have the greatest maximal capacity to produce a moment about the knee joint (Delp et al., 1990). Using the JRS equation, Derouin and Potvin (2005) determined the individual contributions of the 13 muscles crossing the knee joint to rotational stability, about the valgus-varus and flexion-extension axis of the knee. They found that the VL and SM have the greatest potential to stabilize the knee joint in squat postures. Therefore, the identification of these muscles which function as key knee joint stabilizers may prove to be useful during high performance training as specific exercises can be prescribed for these muscles to enhance the promotion for stability, or alternatively exercises can be prescribed during a rehabilitation protocol to aid in recovery. The stability model used by Derouin & Potvin (2005) offers great insight into the stabilizing potential of the individual knee joint muscles.
Chapter 3
Methodology:

3.1 Participants:

Twenty-eight healthy male and female participants (14 female & 14 male), between the ages 19-26 years (females 21.65 ± 1.60, males 22.79 ± 1.48) and with no history of pain or injury in their lower extremity for the previous six months, were recruited from the university population. Both males and females were used in this study, as there is a hypothesized difference between jump landing knee joint mechanics between sexes. Each participant was required to fill out the Lower Limb Health Questionnaire, designed by the researcher, before beginning data collection. Also, participants were required to have an H/Q > 0.45, as measured before participation on the Biodex. Participants were also asked if they are allergic to any adhesives, tapes or rubbing alcohol. Before participating in this study, participants were provided with a detailed description of the methodology, and were required to fill out a letter of consent (attached below).

3.2 Instrumentation and Data Acquisition:

Motions of major joints throughout the entire body, with a higher emphasis on the joints of the lower body, were captured using a 14-camera motion capture passive marker system (Motion Analysis Corporation, Santa Rosa, California) and sampled at a rate of 60 Hz. The marker set consisted of 45 markers (Figure 6). Additionally, eight channels of sEMG were recorded adapting the placement protocol previously utilized by Cashaback & Potvin (2012), with the addition of the hip adductor gracilis. The following muscles were recorded on the participants’
dominant leg: RF, VL, VM, BF, ST, GR, GL, GM. Wireless surface electrodes (Delsys Trigno 4 channel Wireless Electrodes, Delsys, Boston, USA) were positioned to each muscles line of action, between the myotendinal junctions and innervation zones. The sEMG signals were amplified using 8 channels on the Bortec AMT-8 system (Bortec Biomedical, Calgary, Canada, 10-1000 Hz, CMRR = 115 dB, gain = 500-1000, input impedance = 10 GΩ) at a sampling rate of 2100 Hz. Parallel 1.2m x 1.2m force plates (Advanced Medical Technologies, Inc, Watertown MA, USA) were synchronously used to record vertical ground reaction impact forces during takeoff and jump landing, and recorded at 2100 Hz. Motion capture, sEMG and force plate data were all simultaneously collected with Cortex Motion Analysis (Cortex, Santa Rosa, CA, USA). sEMG and force plate data were simultaneously collected at 2100 Hz, while motion capture was recorded through motion analysis Raptor-4 Digital camera’s at 60 Hz.
Figure 6: Retro-reflective marker set as it was applied to participants for the motion capture. There were 45 motion capture retro-reflective markers.
3.3 Experimental Procedures and Protocol:

Participants were required to attend two separate sessions, with a minimum of 48-hours between sessions. The first session was treated as an orientation session, and the second session was treated as the testing session.

3.3.1 Orientation Session:

Each participant’s height (m) and weight (N) were measured and recorded. The experimenter records weight of the participant by having the participant stand still on the force plate for 3 seconds, and height was measured through the use of a conventional measuring tape. The participants were seated in the Biodex (Biodex Inc., Quebec, Canada), and seat adjustments were made to allow the hip, knee and ankle to be held at 90 degrees. The maximum strength of the hamstrings and quadriceps muscles were tested on the participants’ dominant leg. Participants were strapped into a padded cuff supporting the upper ankle. Participants were instructed to complete four repetitions of isokinetic maximal flexion of the knee, followed by maximal isokinetic extension of the knee. From this, the H/Q ratio of the participant was determined.

3.3.2 Testing Session:

Upon arrival to the lab, participants were required to complete a five-minute warm up on a cycle ergometer. Following warm up, participants were instructed to complete three double leg maximal vertical jumps, with the right and left leg landing on the right and left force plate, respectively. A velocity potentiometer was attached to the participants’ shorts, to precisely measure the change in jump height, from neutral to maximum. These values were used as a baseline for the
participants maximal jump height during the progressive fatiguing protocol. After electrode placement, participants lay prone on a mat and a sEMG noise trial was collected for 20 s. While still lying, isometric maximal voluntary exertions (MVE) for each muscle group were collected from the knee joint flexors, extensors and gracilis, and MVE’s of the knee joint flexors were collected from plantar flexing at the ankle. Each of these efforts was performed for 3 s three times in succession, with 30 s of rest after each MVE, with resistance applied by the researcher. Following the MVEs, participants were fitted with 45 retro-reflective markers (Figure 6).

Following calibration of the Cortex system, by taking the participant through static and a range of motion trials, participants commenced the progressive fatiguing protocol. Participants completed double leg maximum vertical jumps every 5 s, with the requirement of taking off and landing with their left leg on the left force plate and their right leg on the right force plate (Figure 7). Participants were given the freedom to adopt a self-selected jumping technique to complete double leg maximal vertical jumps. Once selected, participants were instructed to maintain the same technique during each jump over the course of the fatiguing protocol. After every 5 seconds (12/minute), a bell from a metronome would ring, indicating that the participant was to perform the next maximal vertical jump, and this continued until the participant had reached a level of perceived maximal muscle fatigue, or ended by the researcher due to safety concerns of the participant during the jump landing phase.
Figure 7: A maximal vertical jump being completed by a participant while motion capture, sEMG, Velocity Potentiometer and force plate data is being collected.

3.4 Data Analysis:

Both the kinematic and ground reaction force data were smoothed using a sixth-order Butterworth low-pass filter with a cutoff frequency of 20 Hz in a custom LabView software (National Instruments, Austin Texas). All sEMG data, including data collected during MVE trials and experimental trials, were conditioned using a low-pass filter (cutoff = 500 Hz, 6th order) and a high-pass filter (cutoff = 20 Hz, 6th order), full wave rectified, and low-pass filtered at 2 Hz (Potvin & Brown, 2005). For each muscle, the highest value from the MVE trials was used to normalize the sEMG data recorded during experimental trials. Alternatively, the peak muscle activation from each muscle, recorded during the protocol and MVE
trials, was used to normalize each muscle's maximum sEMG signal. Specifically, all sEMG data, recorded from an individual muscle, were concatenated and a 1-second moving average determined the recorded highest average from that muscle. This process was completed for all 8 muscles. The percent sEMG activation of each muscle being recorded was analyzed individually to observe the effect of continuous maximal jumping on each of the 8 muscles throughout the entire trial. This was observed throughout the trial as fatigue progressed, examining the percent change in muscle activation of each individual muscle throughout the %trial during both pre-takeoff and pre-landing phases of the vertical jump. sEMG was measured during the pre-takeoff phase (instant of take off minus 50 ms) and the pre-landing phase (instant of landing minus 50 ms) of the vertical jump, and was examined throughout the entire protocol to monitor how muscle activations involved in jump landings change as fatigue progresses.

Knee joint angles were measured by analyzing the motion of the retro-reflective markers at the landing phase of the vertical jumps throughout the protocol as fatigue progressed. The landing phase of the knee joint angle was defined as the change in angle between the first ground contact on the return to the force plate during landing, and ending at the bottom of the crouch, when CofG velocity is equal to 0.0 m/s. At this point, participants experienced the greatest angle of knee flexion before beginning to re-extend their knee joint to return their rested standing position.

A variable, termed ‘Effective Duration’, was calculated as the impulse normalized to the peak force during the landing phase (Figure 9). Effective
Duration provides insight into how the participant distributed the reduction in downward momentum, during the course of the landing phase. This was monitored through the use of two force plates, examining the landing impact forces throughout the duration of impact, as well as examining the duration of the jump, which was integrated to approximate the maximum height of the jump. From this, it was possible to see how landing impacts were affected throughout progressive fatigue, as compared to a change in jump height. Figure 8 provides an outline to the location on curve where the dependent variables were evaluated throughout the phase of the jump.

To average data across participants, as fatigue progresses differently for each participant, the data were time-normalized based on number of jumps, using a method referred to as ‘rubber-banding’. This method, adapted from Winter (1980), uses a second order polynomial equation to divide each participant’s jumps into 6 data points representing time zero and all 20% increments of this base time to 100% (Figure 10). This procedure allowed for curves to be averaged across jumps and across participants as fatigue progressed. All participants were normalized to a 100% level, averaging data points between their recorded jumps, to equally analyze and compare all participants on the same 6 points as fatigue progressed. These 6 points are referred to as %trial (0, 20, 40, 60, 80 and 100%) throughout the document.
Figure 8: Example ground reaction force curve, outlining the time points that the dependent variables were evaluated throughout the phase of the jump.
Figure 9: Force-time history showing the effective duration from a maximal vertical jump landing, determined to start at first ground contact and end at the bottom of the crouch (CofG velocity = 0 m/s). The participant in this example weighed 850 N. Impulse, represented by the shaded blue area under the force curve, was determined by calculating the average of all the force data (1564.97 N) and multiplying that by the total duration of the landing impact (t = 0.361 s). The shaded box represents the time in which the peak force of the jump would need to be exerted to equal the total impulse of the entire landing curve (impulse/peak force).
Figure 10(a): This rubber-banding example represents schematic data of maximum vertical jump height of 2 participants. The diamond points represent participant 1, whereas the square points represent participant 2, as each participant progressed through the fatigue trial. The lines represent a fitted second order polynomial curves fit to the respective decline in jump height of each participant respectively over time. The y-intercept would represent the value at t = 0.0 s. For participant 1 (blue bars), a total of 6 intervals were equally spaced at (%trial = 0%), jump = 5.2 (%trial = 20%), 10.4 (%trial = 40%), 15.6 (%trial = 60%), 20.8 (%trial = 80%), and 26 (%trial = 100%), and compared with participant 2 (red bars), intervals from jump 1 (%trial = 0%), 4.4 (%trial = 20%), 8.8 (%trial = 40%), 13.2 (%trial = 60%), 17.6 (%trial = 80%) and 22 (%trial = 100%). Figure 9(b) represents the data points, and how rubber banding aligns time-history between participants, as adapted from the second order polynomial curves, for both participants from Figure 9(a). With the time-history aligned, participants were equally compared amongst each other.
3.5 Statistical Analysis:

A 2x6 analysis of variance (ANOVA) with repeated measures was used to determine the influence of each of the two independent variables: Sex (M, F) and %Trial (0%, 20%, 40%, 60%, 80%, 100%). The significance level for the main and interaction effects of each ANOVA was set at $p < 0.05$.

The dependent variables in this study are: sEMG amplitude, knee angle at impact and effective duration during impact. Omega squared ($\omega^2$) analyses were performed on each main effect and interaction to determine if the explained variance, of any significant main or interaction effects, meets the threshold of $>1\%$ to be considered functionally relevant. In conjunction with ANOVA, if main or interaction effects were proven significant and results met $\omega^2$, a Tukey’s HSD post hoc test was run. Tukey’s HSD post hoc test was performed on any significant main effect of time to fatigue, or interactions between time and sex.
Chapter 4
Results

4.1 Kinetic Data

4.1.1 Total Jumps Completed
A t-test showed that there was no statistically significant difference in the number of jumps completed between sexes (males = 112, females = 100, F=0.626, p=0.436).

4.1.2 Peak Landing Impact Force
Main effects of time (F= 3.8, p=0.032, Figure 11) and sex (F=10.3, p<0.001, Figure 12) were found for peak landing impact force. Post hoc tests revealed statistically significant increases in landing impact force between the time points of 100% and 40% (9% increase), 60% (8%) and 80% (5%). There was a significant increase in landing impact force from time 60 to 80% (3%). In addition, the sex main effect showed that, when normalized to body mass, males landed with a force that was an average of 34% higher than than females.

4.1.3 Jump Height
There were significant main effects of time (F=9.6, p=0.004, Figure 13) and sex (F=23.2, p<0.001, Figure 14) on jump height normalized to the body mass. Post hoc analysis revealed significant decreases in jump height between time periods 0-80% (12%), 0-100% (9%), 20-60% (6%), 20-80% (7%) and 40-60% (2% increase). Males jumped on average 42% higher than females.
Figure 11: Peak landing force as a function of trial time, normalized to each participant's body mass. Means and standard deviations are shown (males n=14, females n=13).
Figure 12: Average normalized ground reaction forces of males and females (males n=14, females n=13). The means and standard deviations are shown.
Figure 13: Main effect of time observed for the decrease in normalized jump height throughout the trial. Mean jump heights are presented along with the standard deviation bars (males n=13, females n=13).
Figure 14: Means and standard deviations are shown for the average normalized jump height comparing males and females (males n=13, females n=13).
4.1.4 Effective Duration

There were main effects of time (F=7.1, p=0.003, Figure 15) and sex (F=5.2, p=0.032, Figure 16) on ED. ED decreased significantly during the following time periods: 20-100% (18%), 40-100% (19%), 40-80% (8%), 60-100% (16%), 60-80% (6%) and 80-100% (10%). The average female ED was 14.5% higher than for males.

4.2 Kinematic Data:

4.2.1 Knee Joint Flexion Angles

There was a main effect of time (F=4.2, p=0.032, Figure 17) on knee joint flexion angle at the instant of jump landing. Knee joint flexion angle decreased at landing between the following time periods: 40-100% (14%), 60-80% (4%), 60-100% (13%) and 80-100% (8%).

4.2.2 Hip Joint Angles

There were no statistically significant main or interaction effects of time or sex on hip rotations about any of the three axes during the instant of jump landing.

4.2.3 Trunk Flexion Angle

There were no significant main or interaction effects of time or sex on trunk flexion angle during the instant of jump landing (p<0.05).
Figure 15: The effect of time on the effective duration following landing from jumps are presented. Means and standard deviation are shown (males n=14, females n=13).
Figure 16: The effects of sex on the effective duration measure for male and female participants. Means and standard deviations are shown (males n=14, females n=13).
Figure 17: The effects of time on knee joint flexion angle for all participants. Standard deviations are shown (males n=14, females n=12):
Figure 18: The effects of time on ankle joint dorsiflexion angle at landing throughout the trial. Standard deviations are shown on the means (males n=14, females n=12)
4.2.4 Ankle Angle

Data revealed that there is a main effect of time during the instant of jump landing on ankle joint flexion (F=9.0, p=0.001, Figure 18). Post hoc analysis revealed significant decreases in ankle flexion angle between the following time periods: 20-100% (12%), 40-80% (6%), 40-100% (14%), 60-80% (4%), 60-100% (12%) and 80-100% (8%).

4.3 EMG

Results of sEMG data will be shown in two sections as described within the methods section: 1) pre-takeoff sEMG activation (50 ms period preceding takeoff, Figure 19) 2) pre-landing sEMG activation (50 ms period preceding landing, Figure 22).

4.3.1 Pre-takeoff sEMG

All muscles, except VM (p=0.391), had a statistically significant main effect of time (p<0.05) on average EMG amplitude pre-takeoff. The following sections will show the results for each muscle. Data for all muscle can be viewed in Figure 19.

4.3.2 Biceps Femoris

There was a significant main effect of time on pre-takeoff BF EMG amplitude (F=9.2, p<0.001). There was a significant decrease in average EMG amplitude at %trial = 40, 60, 80 and 100%, relative to %trial = 0 and 20%, and at %trial = 60 and 80% relative to %trial = 40%.
4.3.3 Gastroc Lateralis

There was a significant interaction effect of time and sex on pre-takeoff GL EMG amplitude (F=9.1, p=0.001, Figure 19). In male participants there was a significant decrease in average EMG amplitude at %trial =60, 80 and 100% when relative to %trial =0, 20 and 40%, and at %trial =80% relative to %trial = 60%.

For females, there was a significant decrease in average EMG amplitude at %trial =40, 60, 80 and 100% relative to %trial =0 and 20%, and at %trial =60 and 80% relative to %trial = 40.

4.3.4 Gastroc Medialis

There was a significant main effect of time on pre-takeoff GM EMG amplitude (F=59.2, p<0.001). There was a significant decrease in average EMG amplitude at %trial = 40, 60, 80 and 100% relative to %trial = 0 and 20%, and at %trial = 60 and 80% when relative to %trial = 40%.

4.3.5 Gracilis

There was a significant main effect of time on pre-takeoff GR EMG amplitude (F=11.2, p<0.001). There was a significant decrease in average EMG amplitude at %trial = 20, 40, 60, 80 and 100% relative to %trial = 0%, and %trial = 40, 60 and 80% relative to %trial = 20%, and at %trial = 60% relative to 40%.

4.3.6 Rectus Femoris

There was a significant main effect of time on pre-takeoff RF EMG amplitude (F=26.5, p<0.001). There was a significant decrease in average EMG amplitude
at \%\text{trial} = 60, 80 \text{ and } 100\% \text{ relative to } \%\text{trial} = 0, 20 \text{ and } 40\%, \text{ and at } \%\text{trial} = 80 \text{ and } 100\% \text{ relative to } \%\text{trial} = 60\%.

4.3.7 Semitendinosus

There was a significant main effect of time on pre-takeoff ST EMG amplitude (F=12.1, p<0.001). There was a significant decrease in average EMG amplitude at \%\text{trial} = 20, 40, 60, 80 \text{ and } 100\% \text{ relative to } \%\text{trial} = 0\%, \%\text{trial} = 40, 60 \text{ and } 80\% \text{ relative to } \%\text{trial} = 20\%, \text{ and } \%\text{trial} = 60\% \text{ relative to } \%\text{trial} = 40\%.

4.3.8 Vastus Lateralis

There was a significant main effect of time on pre-takeoff VL EMG amplitude (F=6.5, p<0.001). There was a significant decrease in average EMG amplitude at \%\text{trial} = 80 \text{ and } 100\% \text{ relative to } \%\text{trial} = 0\%, \%\text{trial} = 60 \text{ and } 80\% \text{ relative to } \%\text{trial} = 20\%, \text{ and } \%\text{trial} = 60\% \text{ relative to } \%\text{trial} = 40\%.

4.3.9 Vastus Medialis:

As noted, there were no statistically significant effects of time or sex on pre-takeoff VM EMG amplitude (p<0.05)
Figure 19: The effect of time on sEMG amplitude on pre-takeoff sEMG. Means are shown for each time point for each channel of sEMG. * interaction effect of time and sex. ** no significant effect of time (males n=14, females n=13).
**Figure 20**: An interaction effect of time and sex on the GL at pre-takeoff showing significant decreases in sEMG amplitude over time. Means and standard deviations are shown (males n=14, females n=13)
Figure 21: Trends of the quadriceps, hamstrings, calf and gracilis muscles shown across the time of the trial at pre-takeoff, revealing a decrease in sEMG amplitude in each muscle group. Means of each muscle group are shown (males n=14, females n=13).
4.4 Pre-Landing sEMG

There was a main effect of time on the pre-landing EMG amplitudes of all muscles observed (p<0.05). There was a main effect of sex on GR (F=27.9, p<0.001), RF (F=6.1, p=0.02), ST (F=9.8, p=0.004) and VL (F=12.4, p=0.002). The VL was the only muscle to have a significant interaction between time and sex (F=4.7, p=0.029). The following sections will show the results for each muscle sEMG amplitudes during the pre-landing phase. Data for all muscle through time can be seen on Figure 2. Post hoc test revealed a series of significant changes in sEMG throughout the time of the trial.

4.4.1 Biceps Femoris

There was a significant main effect of time on BF EMG amplitude pre-landing (F=6.9, p=0.004). However, none of the post hoc means comparisons were significant.

4.4.2 Gastroc Lateralis

There was a significant main effect of time on pre-landing GL EMG amplitude (F=15.2, p<0.001). There was a significant decrease in average EMG amplitude at %trial = 20, 40, 60, 80 and 100% relative to %trial = 0%, at %trial = 40, 60 and 80% relative to %trial = 20%, and at %trial 60% relative to %trial 40%.

4.4.3 Gastroc Medialis

There was a significant main effect of time on GM EMG amplitude pre-landing (F=7.6, p=0.005). There was a significant decrease in average EMG amplitude at
%trial = 60 and 80% relative to %trial =0%, and at %trial = 40 and 60% relative to %trial = 40%.

4.4.4 Gracilis

There was a significant main effect of time on pre-landing GR EMG amplitude (F=8.5, p=0.003). However, there were no significant post hoc comparisons.

There was a significant main effect of sex on pre-landing GR EMG amplitude (F=27.9, p<0.001), revealing that females activate the GR, on average 59%, more than males (Figure 22).

4.4.5 Rectus Femoris

There was a significant main effect of time on pre-landing RF EMG amplitude (F=28.2, p<0.001). There was a significant decrease in average EMG amplitude at %trial = 20, 40, 60, 80 and 100% relative to %trial = 0%, at %trial = 40, 60 and 80% relative to %trial = 20%, and at 60% relative to %trial 40%.

There was a significant main effect of sex on pre-landing RF EMG amplitude (F=6.1, p=0.02), revealing that females activate the RF on average 46% more than males.

4.4.6 Semitendinosus

There was a significant main effect of time on pre-landing ST EMG amplitude (F=4.7, p=0.023). However, there were no significant post hoc comparisons.

There was a significant main effect of time on pre-landing GR EMG amplitude (F=9.8, p=0.004), with females ST EMG amplitudes being, on average, 2% more than males (Figure 23).
4.4.7 Vastus Lateralis

There was a significant interaction effect between time and sex on pre-landing VL EMG amplitude ($F=4.7$, $p=0.029$, Figure 24). There was a significant decrease in average EMG amplitude in males at $%\text{trial} = 20$, 40, 60 and 80% relative to $%\text{trial} = 0\%$, and at $%\text{trial} = 40$ and 60% relative to $%\text{trial} = 20\%$.

There was a significant decrease in average EMG amplitude in females at $%\text{trial} = 40$, 60, 80 and 100% relative to $%\text{trial} = 0$ and 20%, at $%\text{trial} = 60$ and 80% relative to $%\text{trial} = 40\%$, at $%\text{trial} = 60$ and 80% relative to $%\text{trial} = 100\%$.

4.4.8 Vastus Medialis

There was a significant main effect of time on pre-landing VM EMG amplitude ($F=17.0$, $p<0.001$). There was a significant decrease in average EMG amplitude at $%\text{trial} = 20$, 40, 60, 80 and 100% relative to $%\text{trial} = 0\%$, at 40, 60 and 80% relative to $%\text{trial} = 20\%$, at $%\text{trial} = 60$ and 80% relative to $%\text{trial} = 40\%$, and finally at $%\text{trial} = 100\%$ relative to $%\text{trial} = 80\%$. 
Figure 22: The effect of trial time on sEMG amplitude during the pre-landing phase. Means are shown for each time point for each channel of sEMG. * interaction effect of time and sex (males n=14, females n=13).
**Figure 23:** The effect of Sex on sEMG amplitude at the pre-landing phase. Means and standard deviations are revealed for the muscles that are significantly different between males and females (males n=14, females n=13).
Figure 24: The interaction effect of time and sex for the VL at the pre-landing phase. Means and standard deviations are shown (males n=14, females n=13).
Figure 25: Trends of the quadriceps, hamstrings, calf and gracilis muscles shown across the time of the trial. Means of each muscle group are shown (males n=14, females n=13).
Chapter 5
Discussion

The current study was designed to examine the effects of continuous maximal two legged vertical jumps on lower body mechanics. To date, no literature exists on the quantification of either kinetic, kinematic or muscle activations during continuous maximal vertical jumping. These effects were captured through the use of force plates, motion capture and sEMG systems. Fourteen males and fourteen females (n=28) completed the orientation and the testing session comprised of continuous maximal vertical jumps until self-determined exhaustion, ending testing, or when the participant decreased their jump height by 30% of their maximal jump. The dependent variables were grouped into three categories: Kinetics, through the measurements of peak landing impact forces, jump height, and effective duration; Kinematics, measured by the maximum joint angles of the trunk, hip, knee and ankle during the jump landing phase; Muscle Activation, measured using sEMG and reported as a percentage of maximum voluntary contractions for the VM, VL, RF, BF, ST, GR, GM, GL during the pre-takeoff and pre-landing phases.

Overall, when we examined the data from the start to the end of the jumps, participants landed with increased peak forces and less knee flexion angle even though there was an overall decrease in jump height, and energy to be absorbed on landing. Based on this work, my data demonstrates that, as participants progressed towards high levels of peripheral and central fatigue, they compensated by altering their landing mechanics to: 1) reserve metabolic
expenditure to promote for the ability to perform the next maximal jump; 2) reduce further fatigue or promote recovery; 3) reduce the discomfort associated with fatigued muscles.

5.1 Kinetic Data

No statistically significant differences were observed when comparing the total quantity of jumps completed by males and females. This is important to note as, when comparing between sexes, any changes that are observed cannot be attributed to a significantly different jump number between males and females, removing this as a covariant. However, it must be noted that males performed on average 12% more jumps than females.

Based on the results from the maximum landing force (normalized to body mass, Figure 26), including ED, max knee joint angle at landing, and max ankle joint at landing, I hypothesized that a learning effect confounded the effect of fatigue in this study. Optimally, with fatigue being a non-factor, landing impact forces would decrease while participants learned to land, to a point where they landed optimally, and the associated peak force would plateau. In terms of fatigue, we would expect to see peak landing forces increase, as the trial progresses, and participants fatigue level increases. When examining the first 40% of jumps, participants landed with progressively less force as participants learned to complete maximum jumps using more efficient and safe landing mechanics. However, for later jumps, participants landed with progressively increasing force that may indicate inferior biomechanical performance, resulting from muscle fatigue. Later muscle fatigue caused participants to land in a more
rigid posture, specifically, a decreased knee joint flexion angle at landing, resulting in a higher peak force. This same trend was reflected in ED, with results showing an increase in ED from the early portions of the trial (0-40% of jump quantity), followed by a decrease in the subsequent jumps. The greatest decrease in ED, 19% (0.128 s to 0.107 s), occurred between 40-100% of the trial, while the total decrease from the start to the end of the participation was 15% (0.123 s – 0.107 s).

The change in ED can be explained with the results from the knee and ankle joint data, as we found that both joints increased in flexion angle during the early portion of the jumps, which is beneficial for reducing landing severity as the energy from the landing is absorbed over a longer time period, with a decrease in joint flexion angle occurring after the 40% quantity of jumps. The knee joint revealed a 7% decrease in joint flexion angle from 0-100% of participant jump quantity, while a 14% decrease was observed from 40-100% of the participants jump quantity. Similarly, an 8% decrease and 14% decrease were observed in ankle flexion between 0 and 100% and 40 and 100% of the jumps, respectively. This learning-fatigue interaction likely occurred due to the recreational athletic population of this study as the act of continuous maximal vertical jumping is not a common exercise in recreational athletics. Based on our results, participants likely learned to improve their performance metabolically and biomechanically during the earlier jumps, before the effects of fatigue began to dominate, becoming more profound than the effects of learning and negatively influencing their performance.
5.1.1 Peak Landing Impact Force:

Males landed with 34% more force, when normalized to body mass, compared to females. These results can be explained by the differences in jump height between sexes, in that males jumped on average 42% higher than females when normalized to body mass. The peak landing impact force differences between sexes can be partially explained by examining the dissimilarities of the potential energy (PE) between the groups and their knee joint angles when landing from the jump, which will be discussed in more detail in the knee joint angle section.

Since PE is calculated as $PE = \text{mass} \times \text{gravity} \times \text{height}$ (centre of mass, as determined by the vertical travel distance achieved at the apex of the jump), males had greater mass (79.6 kg ± 11.5) and jumped to a greater height (37.8 cm ± 7.9) resulting in a $PE = 79.6 \text{ kg} \times 9.81 \text{ m/s}^2 \times 0.378 \text{ m} = 295 \text{ Joules}$, whereas females (mass 58.3 kg ± 5.6, jump height 19.5 cm ± 5.9) $PE = 58.3 \text{ kg} \times 9.81 \text{ m/s}^2 \times 0.195 = 112 \text{ Joules}$. The importance is not simply that there was a difference in PE but that males and females landed with the same knee joint angle, implying that even though males possessed greater PE, males chose not to adapt greater knee flexion during landing to aid in the dissipation of the greater energy. When examining this from a biomechanical perspective, this result was unexpected, as the males did not choose a method to promote for a “softer” landing, however we hypothesize that the males chose not to increase the knee flexion angle to conserve metabolic energy, as a method to reduce muscle fatigue and discomfort. As fatigue progressed, in an attempt to reserve energy to complete the upcoming jump, participants would sacrifice joint mechanics and land in a more rigid posture with a less flexed knee joint, to
decrease the exertion time (minimizing eccentric work, ultimately reducing overall fatigue and eccentric based muscle damage) and increase recovery time to complete the next maximal effort.

Ultimately, the largest increase in normalized peak landing impact force as observed as a main effect of time, independent of sex, 9%, was seen from 40-100% of the trial. Previous vertical jump landing research completed by Borotikar et al. (2008), and McLean et al. (2005), which induced fatigue in a non-progressive manner, supports the findings of increased peak landing forces while landing in a fatigued state. Additionally, we revealed similar findings to Borotikar et al., (2008) and McLean et al., (2005), in that a decrease in knee joint flexion angle at landing is occurring as jump quantity increases throughout the trial and the participant becomes fatigued, causing a more rigid landing, with the participant decreasing the time spent to absorb the energy of the jump landing. This is the result of landing in a more straight legged and rigid posture, which causes an increase in impact force.
Figure 26: Fictional data showing the hypothesized results from this study, revealing a hypothesized learning curve and a hypothesized fatigue curve. The largest changes in performance hypothesized to be due to fatigue are seen between %trial = 40-100%, rather than from % trial = 0-100% due to the hypothesized effect of learning.
5.1.2 Jump Height

As noted previously, when normalized to body mass, males jumped 42% higher than female participants. This finding indicates that males were able to generate significantly more relative power to overcome their inertia allowing them to thrust themselves higher, when compared to the lighter females. Males tend to have a greater amount of muscle mass, and females tend to have a higher percentage of body fat, which results in greater force being generated per/kg in males, as the physiological cross sectional area in males has a greater percentage of muscle mass, which helps to explain this difference in jump height normalized to body mass (Maughan et al., 1982). Thus, each muscle has a cross sectional area related to muscle force production potential (N/cm²), at an equal body mass, males will produce more force then females per cm², as males have a larger CSA per body mass which will allow them to overcome more physical mass. This is an interesting finding, such that males have a normalized jump that is 42% higher than females, but only results in a normalized landing force that is 34% higher than females. Proportionally through the comparison of jump height to landing force, this is revealing, as females are landing with more force proportionally than males, with respect to their jump height, which may reveal a reason that females are more prone to non-contact knee joint injuries at jump landing.

From the start to the end of the trials, our participants decreased jump height by an average of 10%, such that a 0.040 cm/kg decrease was recorded. The decreased jump height experienced in this study was expected, as Chappell et al., (2005) experienced a similar decrease in jump height in fatigued participants of 0.05 cm/kg from the rested compared to the fatigued height value.
Ultimately, progressive muscle fatigue and central fatigue (described in detail later in this section), resulted in a hypothesized decreasing muscle effort causing the participant to ultimately decrease the ability to thrust themselves vertically, thus causing a 10% decrease in vertical jump height as the trial progressed.

5.1.3 Effective Duration

Throughout the study it was observed that when females landed they did so with an ED that was 14.5% greater than males. A higher ED represents the time the peak forces from the jump landing are dispersed throughout the tissues of the lower body. Therefore, a higher ED indicates that the body absorbed the energy over a longer duration, which is assumed to be beneficial in injury prevention.

The difference observed between males and females in our study is hypothesized to be due to the peak landing force of females being 34% lower than that of males, as statistically, there were no significant differences between landing kinematics between sexes. As effective duration is a calculated result of force impulse divided by peak force, a higher ED will be associated with a lower peak force, if the impulse remains the same. As per our initial hypotheses that ED would decrease as jump quantity increased, a statistical significance was revealed supporting that our hypothesis can be accepted. However, there was no interaction between sex and time revealing that the ED of males and females decreases equally, causing the rejection of our hypotheses that males and females would decrease equally. We have rationalized this finding based on the fact that both sexes maintained a similar biomechanical strategy; specifically that both used a similar knee joint angle by both landing with more extended knee
joint angles as the %trial increased. Male participants had a larger peak force and impulse, yet employed similar knee joint rotations as the female participants. As previously stated, these kinematics would affect ED, as both males and females landed with a more extended knee joint as fatigue progressed, however female participants have less kinetic energy then males to dissipate at landing, and are landing with lower normalized maximal peak forces. As males and females are landing throughout the trial with an equal decrease in knee and ankle joint flexion, and equal rigidity at landing, we can conclude that the higher landing force by males at landing is the cause of the lower ED in males.

Additional evaluation of the ED and lower limb joint angle data provided an addition explanation for the differences between sexes. The higher female ED may be explained by the fact that females landed with greater ankle flexion (commonly referred to as dorsiflexion). Given that there was no difference found in knee angle, the increased ankle flexion may have increased the duration of the landing forces, ultimately decreasing the force at landing, allowing for the human tissues to absorb the landing through eccentric contractions of the plantar flexors muscle caused due to rotation about the ankle. In addition to the ankle, females landed with more hip flexion, which although was not proven to be statistically significant, would decrease the impact and severity of the peak forces. Yeow et al., (2011) support this idea through their examination of pre-post-test performance during single and double leg landings where they found that the main energy absorbers for the landings are the knee and the hip joint. Yeow et al., (2011) findings help to support that the increased flexion of the hip by females
during landing, ultimately assist in the energy dissipation, which could have resulted in the decreased ED observed in female participants.

5.2 Kinematic Data:

5.2.1 Knee Joint Angles:

This study revealed continuous maximal vertical jump caused the participants to alter their biomechanics such that when they landed they did so with a more extended knee joint. Specifically, as participants progressed through their jumps from 40% up to the point where they ceased jumping (100%) they reduced their knee joint flexion angle from a more flexed angle of 77° to a more extended angle of 67°, a 14% decrease. These findings were very similar to those of Chappell et al., (2005) who examined the effects of fatigue on a male and female recreational athletic population in a 5 trial stop-jump tasks and also found significant decrease of 14% of knee joint flexion angle upon the completion of jumps in a fatigue state. Despite our original hypothesis that, as fatigue progressed, females would decrease knee joint flexion angles to a greater extent than males, no significant differences were found (males 72.7° and females 73.7°). In fact, independent of sex, knee joint flexion angle decreased 14% from 40-100% (77° to 67°) revealing that, as fatigue increased, participants landed more rigidly in attempt to reduce metabolic cost. This is interesting, as current literature evaluating the knee joint in a pre-test post-test study method, such as Borotikar et al. (2008), Chappell et al., (2005) and McLean et al., (2005), all revealed that females landed with a
significantly more extended leg than males (decrease in knee joint flexion angle) during the post-test protocol. Contrary to previous findings, our study reveals that progressively completing a single task to fatigue causes both men and women to land with equally extended knee angles when compared throughout the course of fatigue. Interestingly, since our findings differed from those found in pre-post-test study designs, we rationalized that completing the same maximal vertical jumping task to fatigue causes a strategy, independent of sex, where humans may minimize metabolic expenditure at the expense of increased impact severity.

5.2.2 Hip Joint Angles

Although there was no statistical significance found for maximum hip joint flexion at landing, it is interesting that females on average landed with 10.5% greater degree of hip flexion than males (63° compared to 57°, Figure 27). As previously stated, this finding could prove to have resulted in a significant difference in the ED between male and female participants. Also of interest, but not statistically significant, is at the 100% of trial, males land with 12° less hip flexion than females. We believe that this occurred, as females landed with a greater degree of ankle dorsiflexion throughout the study, and flexed their hips to accommodate the forward lean associated with the centre of mass moving forward over the toes. This strategy was likely used to maintain their whole-body balance. Laughlin et al., (2011), in an evaluation of jump landing techniques during soft and hard landings, observed that knee and hip flexion angles were increased when people were instructed to complete a soft landing, and decreased in a more rigid, hard landing. This is an interesting finding in that, although the knee joints remained
the same, the degree of flexion in hip and ankle joints in females actually reveals that females were likely to be landing softer than males.

Hip internal rotation was not statistically significant, resulting in the rejection of the hypotheses that both males and females would land with greater IR as they became fatigued. Throughout the trial, the average internal rotation decreased from 3.2 to 2.1° and, on average, females landed with greater IR (4.1°) compared to males (2.5°). This contradicts the findings of Rodacki et al., (2002) and Borotikar et al., (2008), who found maximally fatigued participants (mostly females), showed significant increase in IR, which is important as such an increase is a risk factor of knee joint injury. However, contrary to those findings, our work showed opposing findings where IR slightly decreased as participants progressed towards fatigue. We have hypothesized that this effect occurred as participants were already beginning to prepare for the next jump while landing from the current landing. It is hypothesized that participants wanted to conserve energy in preparation for the next maximal jump. A lack of IR at the hip joint could be a method used to conserve energy, as the body will not have to recruit muscles to correct for the IR before the next jump is completed.

Finally, statistics completed on hip adduction angle also revealed no statistical significance, with an increase from -7.4 to -6.7° from 0-100% of the trial. This finding bears interest as we originally hypothesized that an increase in knee valgus angle at landing would occur due to a change in hip abduction/adduction angle. However, in the current study, no such changes were observed. We hypothesize that completing a single-task until fatigue affects the mechanical strategy used when landing from a jump in a manner such that,
participants attempted to reduce further fatigue while promoting for muscle recovery for the next jump by not creating large joint angle displacements, thus creating a “softer” landing and ultimately a lower centre of mass. Yet, the strategy observed used less flexion about the joint, which would create for a more metabolically efficient scenario as they would not have to overcome a greater centre of mass displacement, reducing fatigue and promoting muscle recovery, to have additional energy for the next jump effort.

5.2.3 Trunk Angle

There were no significant effects of time or sex on maximum trunk flexion angle, in fact there was only a negligible decrease from 0-100% of the trial, and a negligible difference between males and females. These findings suggest that either the angle of the trunk must remain constant in order to successfully complete a jump landing, or, that altering the trunk angle in either flexion or extension would require muscle effort to counteract such motion. This is a relevant finding in that, maintaining a neutral trunk angle throughout the duration of the jumps is critical in reserving energy, as a non-neutral trunk angle (flexion or extension) upon landing would require wasted metabolic energy, as too much trunk flexion would require the participant to step forward to land adequately, and if the trunk was too extended, then the participant would have to step backwards. Therefore, maintaining a more neutral trunk angle reserves metabolic cost, in that the jump landing technique remains in a good balance, reserving energy to prepare for the next maximal jump.
5.2.4 Ankle Angle

There was no significant effect of sex on the ankle angle during landing, however it is interesting to note that females, on average, landed with 15% greater ankle flexion than males throughout the trial (Figure 27). As previously mentioned, we hypothesize that a greater degree of ankle dorsiflexion may have an effect on the ED as this resulted in a longer duration of landing, ultimately revealing a softer landing in female participants. It is interesting in that our results tend to show that females appear to respond to fatigue by landing with more biomechanical benefits than males, even though previous works suggested the opposite would occur. Based on our data, it appears that females, land with the same degree of knee flexion as males, however utilize a mechanical strategy that may dissipate force through more hip and ankle flexion during fatigued jump landings.
Figure 27: A comparison of the final (%trial = 100%) average landing posture of males and females. As observed, males land with decreased hip and ankle flexion compared to females. Knee and trunk angle at landing were equal between sexes.

5.3 sEMG

The electromyography (sEMG) data contributed very interesting results providing significant knowledge of both pre-takeoff and pre-landing time periods. These results will be explained in detail for each muscle below. It is important to mention that sEMG was not evaluated at jump-landing, as intense impacts during a jump
landing resulted in large amount of motion artifact of the sEMG electrodes due to the skin movement (as viewed in slow motion video).

When examining both the pre-takeoff and pre-landing phases of each jump, throughout the duration of the entire trial, a decrease in sEMG amplitude was found for all muscles. The observed decrease is hypothesized to have occurred due to both a combination of central and peripheral fatigue that are explained in the following paragraphs. Our explanation begins with the peripheral fatigue. We hypothesize that the decreased sEMG activation found in all muscles was partially due to a reduction in available motor unit’s (MU) to produce force, which occurs when the MU’s reaches their exhausted state and thus, will shut off entirely which results in an overall decreasing amplitude of the recorded sEMG (Cifrek et al., 2010, De Luca et al., 1979). We observed this MU exhaustion in our study based on a decrease in both jump height and sEMG amplitude as %trial increased towards the end of participation. It must be noted that an interesting event occurred as there was a slight increase in jump height (0.007 cm/kg) at the end of the trial (%trial = 100%), compared to the %trial = 80%. This was unexpected especially for the quadriceps muscles, the muscles most responsible for vertical jumping, which showed a 26% decrease throughout the trial from 0-100%. However, further examination showed that the calf muscles increased their sEMG amplitude by 2% from %trial = 80% to %trial = 100%. We have hypothesized that due to the reduction in quadriceps sEMG, the calf muscle, in the 80-100% of trial, increased activation slightly to aid in the vertical jump forces during the time from 80 to 100 %trial and thus, the slight increase observed in force was due to the increased activity of the calf muscles. Additionally, we
acknowledge that sEMG was not recorded from all leg muscles and, thus, this increase in jump height may have been a result from their increased efforts.

The decrease in sEMG may also be due to central factors, such as participant motivation. Central factors in the study of muscle fatigue are those physiological processes that occur within the CNS, including the ability to generate a sufficient and appropriate command to complete a task (Enoka, 1995). Although no central fatiguing measures were recorded in this current study, it is important to address the potential role that central fatigue had on this task. Kent-Braun (1998) in a study examining the effects of central and peripheral fatigue, reported that central fatigue could account for up to 20% of the muscle fatigue developed during maximal efforts. If participant motivation became a factor in this study, the effect of decreasing sEMG activation would affect, either consciously or unconsciously, the recruitment of MU’s to complete the maximal effort, with a decreasing motivation correlating to a decreased MU recruitment. Presumably, a participant with a decreased motivation, or a decreased tolerance to pain, would decrease contraction strength and effort throughout the trial as a means to reduce fatigue-induced discomfort (Enoka, 1995). For a participant to sustain activity while completing a fatiguing task, a central command signal must consistently produce a signal that allows all muscles to function at their optimal performance level, of all the muscles involved in the task. Fatiguing of this central system would correlate directly with a decreased sEMG signal, as observed in this current study, as when the signal command centre begins to fatigue, the centrally produced signals are decreased from those in a rested situation. Additionally, a theory that is outlined by Revill ad Fuglevand (2011), suggest that
an effect of excitation adaption can occur within the motor units, from a time-dependent reduction in firing rate in response to a constant current input. In the current study this is represented by a continuous maximal jump protocol, and as fatigue progresses, is characterized by a slow exponential drop in firing rate proportional to the magnitude decreasing firing rate by as much as 40–60%.

When comparing our results to previously related works we found conflicting sEMG results. Specifically, Rodacki et al., (2001) examined jump tasks in a pre-post-test method, with fatigue being induced by leg extensions in a weight machine, found significant increases in knee joint muscle sEMG activation when comparing pre to post-test measurement periods. For example, Rodacki et al., (2001) showed that after the fatiguing level, sEMG levels of the quadriceps were up to 39% greater than the resting levels, and the hamstrings were up to 29.5% greater during fatigue then rest. However, that study differs from ours in that our participants maximally jumped to a fatigue state, whereas Rodacki et al. (2001) completed submaximal efforts, and fatigued their participants using a common leg extension exercise. The results from Rodacki et al. (2001) are interesting, in that the sEMG recording of the quadriceps muscles VL and RF, the power generating muscles for the action of maximal vertical jumping, increased in activation by 39%, while the hamstring muscle ST increased 18.8% during the fatigue test.

Based on all of the sEMG results, we hypothesize that a combination of peripheral fatigue - the physiological effect on the muscles from producing force to move the body in a maximal effort - and central fatigue, the motivational levels,
as well as CNS inhibition due to fatigue, of the athlete to continuously complete a challenging protocol, had an effect on the sEMG in this current study. To support this explanation, Bigland-Ritchie et al., (1983) shows the effects on EMG in a sustained maximal contraction of the adductor pollicis muscle. Their results revealed that sustaining a maximal contraction for 60 seconds showed a decrease in sEMG amplitude, due to what they believe was a large effect of central factor and neural drive, as the force did not decrease as significantly as the sEMG.

When examining all of the sEMG data, we gain a greater understanding of the muscle behaviours based on their different roles throughout a jump task. For instance, based on the biomechanics of jumping, pre-takeoff muscle activation was expected to be the highest in the quadriceps and gastrocnemius muscles, as these are the muscle mostly responsible concentrically contracting to produce joint extensor moments to vertically thrust the body during a jumping task. This muscle behaviour was evident, as the activation of these muscles during pre-takeoff were substantially higher than the activation of the hamstring and gracilis muscles, which act more as joint stabilizers during this phase of the jump.

In our study, pre-landing values were viewed as the preparation for landing. This allowed us to determine if participants were beginning to contract proactively in anticipation of the jump landing. It is not surprising that, throughout the trial, we observed decreases in sEMG amplitude in all muscle groups, as %trial increased to the fatigue state. Based on this decrease, we hypothesized that the airtime was used subconsciously as a time for the muscles to rest and recover, knowing that they will have to again complete a maximum jump effort 5 seconds following
the landing. It must be noted that a change in values from the pre-landing sEMG activation from %trial = 0-100% are very low, and are recognized by the researcher to be of potentially little significance, and could potentially represent a very minor change across the %trial, which may not be explained solely by fatigue that occurs due to the continuous jumps during the trial.

When examining the knee flexor BF, the sEMG at the pre-takeoff and pre-landing phases decreased from %trial = 0-100% by 25% and 32%, respectively. The BF contracts eccentrically during the jump propulsion phase, however the BF is important during jump landing, stabilizing the knee during impact, so that the knee does not buckle. The BF is constantly required to produce force during jump landing, which fatigues the muscle, ultimately reducing its ability to continually produce force as jumps progress.

Like the BF, the ST has the role of knee flexor and thus, contracts eccentrically during the jump phase. However, during landing, the ST concentrically contracts, which plays an important role, like the BF, in providing flexion about the knee stabilizing the knee during impact, so that the knee does not buckle. The BF experiences significant decreases in sEMG from 0-100% during the pre-takeoff phase with the activation decreasing 17%, while during the pre-landing phase sEMG decreased 34%. However, the BF has a greater force producing potential and due to this, is responsible for a much larger percentage of force than the ST. Also, as previously noted, we only measured two of the hamstring muscles, and thus cannot expand these thoughts to other muscle such as the semimembranosus. The hamstring muscles throughout the phases of jumping and jump landing play an important role in stabilizing the knee joint, by
optimally producing forces for stabilization while minimizing any excessive knee flexion forces. This delicate strategy of stabilizing the knee while not requiring greater quadriceps forces to overcome any unnecessary knee flexion motion about the knee is optimal for both jumping and landing phases. As for the quadriceps, they concentrically contract to produce the force needed to maximally jump, and then eccentrically contract during jump landing to cause a deceleration of the body. To complete the deceleration, the quadriceps are used to complete negative work that is used to reduce the kinetic energy, that is created as the body moves towards the ground, to a value of zero, which occurs at the bottom of the crouch, and once this is achieved the body has ceased the landing phase.

The VL, the main force producer of the quadriceps muscle group, revealed a decrease of 14% from 0-100% of the trial during the pre-takeoff phase. The quadriceps muscle group (VL, VM and RF measured in this current study), along with the gastrocnemius muscles (GL, GM) provides the majority of the muscle force to complete the takeoff proportion of a vertical jump, while the hip extensors (ST, BF) play a role in stabilization of the knee joint throughout the jump-landing. These muscles contract concentrically in order to drive the body vertically during the maximal vertical jump. According to Friederich and Brand, (1990), Brand et al., (1986), and Wiekiewicz et al., (1983), as adapted from Cort and Potvin (2011), the force potential of the VL is the greatest of all the quadriceps group, and the GM provides 2.5 times the amount of force compared to GL. Interestingly, our data reveals the VM, which shares the same role as the VL to
propel the body vertically, did not show significant sEMG decreases throughout the trial. The values of the VM remained constant throughout the trial, which leads us to believe that the VL, due to its force potential and sEMG magnitudes, had the greatest responsibility in the vertical jump. Furthermore, the RF revealed significant decreases of 21% from time 0-100% (76% to 63% activation) in the pre-takeoff phase. This is also interesting in that the RF, has a lower force generating potential than both the VL and VM, and, based on our results, fatigued at a similar rate as the VL. These findings may reveal that, during a same-task fatiguing protocol, the VL and RF are utilized the most, while the VM sEMG is unchanged, to reserve energy to continue to complete the same task continuously. Additionally, the VM may not have ever reached its maximum force, activating fast twitch fibres, such that its fatigue would differ from other muscles, as the slow twitch fibres remained at a relatively constant activation level throughout the trial, without the need to recruit the fast twitch fibres.

The GL muscle, a muscle that when contracted concentrically helps in producing upward force during a maximal vertical jump, decreased 26% from 0-100% of the trial. Much like the GL, the GM showed a significant decrease in activation of 26% during the pre-takeoff phase. Both the GL and GM had an identical decrease of 26% at the pre-takeoff phase of the trial, over the course of the jumps, revealing the reliance on the calf muscles of both concentric and eccentric loading during a continuous maximal vertical jumping task.

Finally, the GR decreased 13% from 0-100% of the trial during the pre-takeoff phase. The gracilis, a small hip adductor, was added to this study to understand if it affects the abduction/adduction angles during landing. This
reduction in gracilis activation may have resulted in the very minimal increase in hip adduction angle we observed at landing, in an attempt to reserve energy for the landing to improve the chances to land with a safe degree of hip adduction and thus, avoiding knee valgus at landing.
5.4 Hypotheses Revisited

4) Females will have greater decreases in overall knee joint flexion angle, and greater increases in knee joint valgus and internal rotation angles, during the landing phase, as fatigue progresses. Fatigue-related changes in knee joint flexion angle will be seen in both sexes, however, females will experience larger changes with time, compared to males. This will be evaluated with a 2x6 mixed ANOVA showing a statistically significant main effect (p<0.05).

The current results failed to reject the null hypothesis, as females and males showed equal decreases in knee joint angle with no significant difference in IR or adduction angle at landing. However, it should be noted that both sexes showed a 10° decrease in knee joint flexion angle at landing when comparing the data from the beginning of the trial to the end.

5) As fatigue progresses, there will be a significant increase in normalized sEMG for vastus lateralis (VL), vastus medialis (VM), biceps femoris (BF), semitendinosus (ST), gastrocnemius lateral (GL), gastrocnemius medial (GM), yet there will be a decrease in rectus femoris (RF). Fatigue-related changes in muscle activation will exist for both sexes (p<0.05), however, post hoc tests will reveal that females experienced larger changes with time, compared to males.

It is important to first note that, since this hypothesis was proposed, recording of the GR was added to the study.

The current study failed to reject the null hypothesis that there would be an increase in normalized sEMG for the VL, VM, BF, ST, GL, GM. However, the null hypothesis is rejected for RF, as significant decreases were seen across the trial in this muscle. Additionally, this study failed to reject the null hypothesis that would reveal females experiencing larger changes with time, compared to that of males, as this is not what was observed in this study.
Additionally, although no central fatiguing factors were measured, we suspect that an interesting combination of central and peripheral fatigue mechanisms resulted in the rejection of this hypothesis.

6) Fatigue, from repetitive jumping, will result in changes in technique resulting in a decrease in effective duration for males and females, as time progresses. However, there will be an interaction effect between sex and time, ultimately revealing that, as fatigue progresses, female participants decrease significantly greater (p<0.05) than males.

This study was successful in rejecting the null hypothesis, in that fatigue from repetitive jumping ultimately changed the landing technique, revealing significant decreases in ED in both sexes as time progressed throughout the duration of the trial. However, this study failed to reject the null hypothesis with regards to interaction effect, as there was no interaction effect between time and sex as fatigue progressed, showing there is no statistical difference in the way that females and males decrease ED through time.

5.6 Limitations and Assumptions

In the current study, the limitations and assumptions that have been identified are related to the jump task learning effect associated with recreational athletes. We acknowledge that completing continuous maximal vertical jumps is not a common exercise completed by recreational athletes and, therefore, this is not a well-practiced skill that can be completed optimally without proper training. An effective jump is defined as the participant reaching maximal height values and landing softly on the plate in an attempt to absorb the forces associated with the
landing with minimal amount of joint rigidity (cushioning the landing with greater knee joint flexion angles, between 77-80°). We underestimated the influence of a jump task learning effect by only providing time for up to 3 maximal vertical jumps pre-testing practice. This amount of practice proved to not be insufficient, as the results of this study reveal that from 0-40% of the trial, participants were actually improving their landing mechanics and jumping performance, before the effects of fatigue began to lead to decrements in their performance. This is based on an observed decrease in peak landing force and an increase in ED, as well as knee and ankle joint flexion angle at landing between %trial = 0-40%. Participants were instructed to perform 3 maximal vertical jumps before the trial for jump training. However, this proved to be insufficient as our results revealed that, at 40% of the trial (~40-50 maximal jumps), participants had continued to improve their jump landing mechanics. The use of untrained recreational athletes, for the task of continuously maximally jumping, may have had an effect on the results of this study, as they may have introduced confounding effects on the data that could skew the interpretation of our data. We believe this to be the case, as previously stated, because the results demonstrate a possible learning effect, for the initial 40% of the trial. With more experience in completing this skill in high performance athletes, it is likely that participants would alter their technique to jump and land more effectively, ultimately decreasing the metabolic expenditure during takeoff and jump landing.

Another limitation of this study is the assumption that a 30% reduction of maximal jump height would indicate participant fatigue, and thus stop the trial.
However, based on our results extreme participant fatigue occurred by the time they had lost 10% of their maximal jump height, and thus our protocol was altered to identify fatigue based on 2 conditions: 1) the participant was experiencing perceived extreme muscle fatigue and they chose to not continue, 2) the researcher ended the trial for the safety of the participant as a noticeable dangerous change in body mechanics was occurring. Although this protocol did not result in equal thresholds for identified fatigue, we chose to ensure the safety of our participants and did not allow them to increase their risk of injury due to high levels of fatigue. It must be noted that, since no publications were found of research involving progressive fatigue during continuous maximal jumping tasks, the magnitude of decreased jump height due to fatigue was much less than what we hypothesized (10% rather than the original 30% proposed), however participants were visually fatigued as determined by the researchers. In fact, it was remarkable that they continued to produce enough force to continuously complete continuous maximal jumps within 10% of their rested maximum.

Finally, this study was designed to examine recreational athletes who actively train. However, the degree of performance and motivation in such athletes varies and unfortunately, such variables were not accounted for. It is believed that the results of this study may not be fully transferable from this population to a high performance athlete population, however additional research would be required to prove this thought.
Chapter 6
Conclusion

This study demonstrated that, as fatigue progresses throughout continuous maximal jump efforts, there was an associated increase in normalized peak landing force. This increase is also significantly different between sexes when normalized to body mass, with males landing on average with 34% greater normalized force than females. Additionally, when normalized to body mass, males jumped and average of 42% higher than females, but both significantly decrease jump height up to 10% as fatigue progresses. Additionally, it was observed that there was no significant difference in the number of jumps completed by males and females to fatigue (males=112, females=100). Further, effective duration significantly decreased as fatigue progressed throughout the trial, by up to 19%, from %trial = 40-100%. Interestingly, it was revealed that females have a 14.5% higher ED than males.

The kinematic data collected in the current study showed that, as fatigue progresses, there was a significant decrease of 10° in the knee joint flexion angle observed at landing. Also at the knee joint, no statistical significance was observed between sexes as fatigue progressed, in the rate at which knee joint angle decreased, or at the resultant knee angle at the end of the participants’ jumps.

At the ankle joint, significant decreases in flexion were observed as fatigue progressed throughout the trial. Also of interest, females landed on average with a greater degree of ankle flexion than males do at landing.
At the hip, although not statistically significant it was observed that females tended to land with a greater degree of hip flexion than males. Additionally, there was no significant changes observed in the hip IR or adduction angle at landing, nor in the trunk flexion angle at landing.

The sEMG data showed that time (%trial) had a main effect on the results recorded from BF, ST, GM, GL, GR, RF, and VL during the pre-takeoff phase of the maximal vertical jump, and BF, ST, GM, GL, GR, RF, VM and VL revealed significant main effect of time during the pre-landing phase of the vertical jump. Trends in the data reveal significant decreases in sEMG amplitude throughout the duration of the trial, as the effects of some combination of central and peripheral fatigue progress to higher levels.

The sEMG data showed that sex had a main effect on the results for GR, RF, ST, and VL during the pre-landing phase. Although the data revealed statistically significant results, they are taken with caution as the change in the values from 0-100% of the trial are very low, and cannot necessarily be directly related to the study intervention.

Finally, the EMG data revealed an interaction effect of time and sex on the GL during the pre-takeoff phase, and the VL during the pre-landing phase, showing significant decreases throughout the trial from 0-100%. The GL effect was more amplified in male participants than females, as males started at 0% with higher sEMG amplitudes, decreasing to the same value as females by the end of the trial. In the VL, males and females revealed equal decreases in sEMG activation over time, however females activate the VL at a higher level than males throughout the pre-landing phase.
6.1 Future Research Directions

Future research should concentrate on providing an adequate training session to the participants who will be completing the continuous maximal vertical jumps. It would be recommended that participants complete no less than 50 maximal continuous vertical jumps, during a separate training session, to adequately learn the task, which will reduce the effect of a training effect seen in our study. Based on our data, 50 jumps are recommended to reduce any learning effect, that at %trial = 40, participants were improving on body mechanics while landing, and only after this did we see a decrease in mechanics, leading us to believe that a minimum of 50 jumps is required to provide a hypothesized adequate amount of maximal jump training to recreational athletes, however, more training may be required.

Additionally, it is recommended that the researcher provide a significant warm up period once the sEMG and motion capture markers applied, to allow the participant time to prepare for exercise, in order to record the most accurate data from completing maximal jumps.

Finally, this study was completed on recreational athletes, where the effects of fatigue are amplified. Future research on high performance athletes would provide better insights on the effect of training on jump landing, which can counter the negative effects of fatigue on body mechanics during jump landing.
Reference List:


Appendices
Appendix A

Today's Date: February 10, 2016
Principal Investigator: Dr. Joel Cort
REB Number: 32993
Research Project Title: REB# 16-015: "Understanding the effects of progressive fatigue on impact landing force and knee joint mechanics, during the landing phase of continuous maximal vertical jump"
Clearance Date: February 10, 2016
Project End Date: December 25, 2016
Milestones: Renewal Due-2016/12/25(Pending)

This is to inform you that the University of Windsor Research Ethics Board (REB), which is organized and operated according to the Tri-Council Policy Statement and the University of Windsor Guidelines for Research Involving Human Subjects, has granted approval to your research project on the date noted above. This approval is valid only until the Project End Date.

A Progress Report or Final Report is due by the date noted above. The REB may ask for monitoring information at some time during the project’s approval period.

During the course of the research, no deviations from, or changes to, the protocol or consent form may be initiated without prior written approval from the REB. Minor change(s) in ongoing studies will be considered when submitted on the Request to Revise form.

Investigators must also report promptly to the REB:
a) changes increasing the risk to the participant(s) and/or affecting significantly the conduct of the study;
b) all adverse and unexpected experiences or events that are both serious and unexpected;
c) new information that may adversely affect the safety of the subjects or the conduct of the study.

Forms for submissions, notifications, or changes are available on the REB website: www.uwindsor.ca/reb. If your data is going to be used for another project, it is necessary to submit another application to the REB.

We wish you every success in your research.

Alan Scoboria, Ph.D.
Chair, Research Ethics Board
2146 Chrysler Hall North
University of Windsor
519-253-3000 ext. 3948
Email: ethics@uwindsor.ca
Appendix B

Understanding the effects of progressive fatigue on impact landing force and knee joint mechanics, during the landing phase of continuous maximal vertical jump

CONSENT TO PARTICIPATE IN RESEARCH

Investigators: Joel Cort, Paul Leuty, Chad Sutherland
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Student /Co-Investigators: Paul Leuty
Chad Sutherland

Purpose of this Research:
The aim of this study is to investigate if impact forces (vertical ground reaction forces) increase with progressive fatigue while body mechanics are sacrificed in landing from a maximal vertical jump. Generally this study aims to investigate whether recreational athletes will sacrifice body mechanics for metabolic costs. Through an examination of kinematic and kinetic data, the aim of the research is to determine the overall effect that progressive fatigue of same-task progression towards fatigue will have on the body mechanics of the individual. You will be required to maximally vertically jump on force plates until a certain level of fatigue is reached.

Procedures Involved in the Research
You will be required to visit the Occupational Simulation and Ergonomics Laboratory at the University of Windsor (room HK209) on two separate occasions, with a minimum of 48-hours between the sessions where, the first session will be treated as an orientation session, and the second session will be the testing session. During the testing session, you will be required to wear clothing that fits tightly to the skin, as for the motion capture and EMG adhesives to remain in the precise location where they are placed. This will be detailed below. The methods are outlined in detail below:

Orientation Session:
1. You were selected for this study, because you are a male or female between the ages of 17-26 years of age. Before participating in this study, you must read this detailed description of the methodology, and at the end, be required to fill out a
letter of consent (attached below), and a lower limb health questionnaire before participation.

2. Orientation & Consent: Investigators will guide you through the consent form, outlining all methodologies, risks, benefit of the research, and most importantly the option to withdraw from the study at any time, without any consequence. At the end of the consent form, you will find a lower-limb health questionnaire which you are required to fill out. An answer of ‘yes’ on this questionnaire will cause you to be ineligible to participate in this study. If you have any questions, at any time, please ask the researcher.

3. Anthropometric data recording: Your height (meters) and weight (Kilograms) will be measured and recorded.

4. Measure Hamstring/Quadriceps ratio on Biodex: You will be seated in the Biodex (common training & exercise equipment which is used in training and rehabilitation), and adjustments will be made to the seat location in order for the knee, hip and ankle to be at 90 degrees. You will be instructed to complete a maximal leg flexion effort (pushing down on the leg press cuff), followed by a maximal leg extension effort (pulling up on the leg press cuff). With this information being recorded by the Biodex machine.

5. After completion of the Biodex protocol, you will schedule your return date to the lab for the testing session, which will be a minimum of 48 hours after the orientation session. For males, athletic shoes that you are comfortable jumping in and tight fitting spandex shorts are recommended. Males will have the option of going shirtless or wearing a tight, spandex, form-fitting shirt. For females, comfortable shoes for jumping, tight fitting spandex shorts, and a sports bra is recommended, with the option of wearing a tight, spandex, form fitting shirt. The reasoning for the clothing is that motion capture markers represent an exact area about their placement location on a participant. Upon placement of motion capture markers, it is imperative to accommodate the need of accurate placement (the motion capture marker can not be moved from the area of initial placement) as much as possible to avoid any extra motion for accurate results, which can be contained by placing the motion capture markers directly on the skin or as close as possible to the skin (tight-fitted clothing) for accuracy in results.

6. This orientation session will take up to 30 minutes

*note: it is important to remember that this research is not in any way associated with your academic standing at the University. Participation is completely voluntary, and the results of this study and your performance will have no affect on any of your grades, or academic standing.

**Testing Session:**

1. You will return to the lab and upon arrival the researcher will sit down with you and go over all the methods involved in the testing session. You will be given time to ask any questions you have regarding the previous orientation session, or
the upcoming testing session. You will then be required to fill out a *returning participant questionnaire* with regards to the current state of your leg muscles to ensure that they have not completed any exercise in the previous 48 hours, and that they are currently not experiencing any leg muscle fatigue or delayed onset muscle soreness as a result of previous exercise exertions.

2. Practice maximum vertical jumping task: You will be given instructions on how to jump as hard and as high as you are capable of with your choice of jumping posture, ensuring that you keep the same posture throughout the duration of the study (for example, if you jump with your arms up, you must continue to jump with your arms up for the duration of the study, for consistency).

3. Also, during the practice of maximal vertical jumping, you will be required to practice landing on separate force plates, with the left foot on the left force plate, and the right foot on the right force plate, as this critical in comparing the changes which occur with fatigue between the left and right leg.

4. Warm-up (5 min stationary cycling): You will then be warm up your lower extremity muscles, in order to increase blood flow & heart rate in preparation for exercise. Using a cycle ergometer you will stationary cycle for a required 5 minutes.

5. Three Maximal vertical jumps (Instructed to perform maximum height): after you are adequately warmed up, you will be asked to complete three maximal jumps in the same manner in which you completed your practice jumps. Jump height will be measured using a velocity potentiometer attached to the back of your shorts. A velocity potentiometer, basically a string which changes lengths, functions to measure the change in length of the string, which represents your jump height; and as the length of the string changes, this will directly measure the change in jump height. This is mandatory in order to determine your individualized baseline measures for their maximal jumping height, as to later monitor the effects of fatigue as you begin the fatiguing protocol.

6. Surface wireless electromyography (sEMG) skin preparation and skin electrode placement: Self-adhesive surface EMG electrodes will be placed on the belly of a series of your muscles. In the anterior compartment of your thigh (Quadriceps), electrodes will be placed on the rectus femoris, vastus lateralis and vastus medialis. In the posterior compartment on your Hamstring, electrodes will be placed on the biceps femoris (long and short head will be a single channel) and semitendinosis (ST). Also on the posterior section, below the knee, there will be an electrode placed on the gastrocnemius lateral and the gastrocnemius medial (Calf muscle, which is comprised of two heads). The location of the muscles will be identified by palpation during maximum isometric contraction pre-testing. To palpate (examine a specific area, using touch), the researcher (Paul Leuty) will require you to flex the muscle group fully, while using two fingers to feel for the center of a specific muscle. Instructions will be given to you as to how you are required to flex. The researcher will speak with you to ensure you are aware of the process, and allow for any questions before beginning. The researcher will demonstrate a palpation on themselves prior to any palpation to you, as to better understand the procedure required in placing sEMG, and eventually, motion capture markers. Your skin may require preparation for the EMG electrode tape...
by means of shaving (if necessary), and disinfection with rubbing alcohol (if necessary) in order for maximum adhesion during the protocol. A new disposable razor will be provided to you. You will be guided by the researcher where you are required to shave. Disposal of the razor after use, will be placed into a Sharp Objects Bin. If you experience a minor cut while shaving, hydrogen peroxide, cotton swabs, and band aids are available in the labs first aid kit, tell the researcher immediately.

7. 3 – 5s MVE (Calves, Quads, Hamstrings); After all the wireless EMG electrodes are placed, you will be required to complete maximal voluntary exertions (MVE) and a noise trial. You will lay prone on a mat and a sEMG quiet trial will be collected for 20 s (sEMG collection while you remain completely still). While still lying, isometric maximal voluntary exertions, contractions which involve pushing on a surface that does not move, (MVE) for each muscle group will be collected from the quadriceps, hamstrings, and calf muscles. Quadriceps will be tested by the researcher opposing the flexion of the your knee, as you maximally contract, by holding the top of your foot as you try to drive their leg upwards (still laying on your back). Alternatively to get the MVE of the hamstrings, the researcher will hold your heel as you attempt to maximally drive your leg downwards extending their leg (still laying in the prone position). MVE’s of the calves will be collected from maximally pointing your toes, at the ankle. Each of these efforts will be performed for 3 s three times in succession, with 30 s of rest after each MVE, with resistance applied by the researcher. The MVE will be used to determine the maximal amount of muscle activity, which is used to normalize the rectified sEMG signal of the trial data.

8. Skin preparation and application of retro-reflective markers; A series of motion capture markers will be placed on your skin with two sided tape in order to complete motion capture for kinematic data. Areas of placement will be defined by palpation, following the same process as sEMG placement, to ensure accuracy. The researcher (Paul Leuty) will use two fingers to palpate bony landmarks about your joints, or limb centers, to ensure accurate placement. The researcher will inform you of all the palpations before they happen, and allow you to ask questions at any time. Retro-reflective markers will be placed over joints, as to accurately document the movements during motion capture. A total of 45 markers will be placed on your skin.

9. Begin Progressive Fatiguing Protocol: you are now prepared to complete the fatiguing protocol. You will be required to stand on the force plates, as previously trained with your left foot on the left force plate and your right foot on the right force plate. You will be instructed to jump as high as you can every 5 seconds, until you are fatigued which is defined as a 30% decrease in jump height. You are required to maintain the same jump technique throughout the duration of this protocol. Once you cannot reach 70% of maximal jump height (a 30% decrease in jump height) as previously monitored, for 3 consecutive trials, the trial will be completed.

10. You will then be assisted in removing the sEMG electrodes and retro-reflective markers, allotted 5-8 minutes of active muscle recovery on the exercise bike any questions will be addressed, and if not, you are okay to leave the lab. You will be reminded that if they are interested in the results of the study, to leave their email
address with the primary investigator, or alternatively, you will be provided with the information associated with attaining the study results anonymously from the University of Windsor’s Research Ethics Board results posting page. http://web4.uwindsor.ca/units/researchEthicsBoard/studyresultforms.nsf/VisitorView?OpenForm&count=-1

11. The testing session is expected to take up to 120 minutes.

WITHDRAWAL:
As a participant in this study, you may freely withdraw from this study at any time. Participation in this study is strictly voluntary and you are free to terminate your participation in this study without any consequence at any time either before or during the testing sessions. Withdrawal is no longer possible once the data is de-identified and amalgamated into a larger data set.

POTENTIAL RISKS AND DISCOMFORTS:
Although limited, the potential risks and discomforts associated with this study are listed below:

Fatigue and muscle soreness – You may have local muscle soreness in your quadriceps/hamstring or calf muscles during, and for several hours after the experiment due to the progressive fatiguing protocol. This muscle soreness, which is typical of any physical exertion study, will be equivalent to a bout of moderately strenuous exercise and should not persist beyond 24 hours. Although muscle soreness is a possibility, it is a very common sensation for people after exerting brief maximal and submaximal efforts. The investigators are qualified to teach the participants muscle specific stretches in order to help relieve the muscle soreness and therefore no medical support is necessary. Participants are free to withdraw from the experiment at any time should they feel excessive discomfort. Also, after the completion of data collections, all participants will be instructed on common stretches in order to reduce discomfort.

Muscle and joint injury - Care will be taken to ensure that the participants do not employ improper techniques that jeopardize the integrity of the tissues. Furthermore, participants will not be required to maintain the exertion or posture for long periods of time, and as they are not actually holding any real weights, they are simply jumping vertically resisting only their own body weight, there is no risk that an object will weigh too much, or be dropped resulting in injury.

Skin irritation - participants will be asked if they have any previous reactions to any adhesive bandages, tapes or rubbing alcohol; if so, they will be asked to withdraw from the study. Additionally, although very rare, some participants may experience a temporary reaction to the adhesive from the surface electrodes. If irritation develops during testing, the instrumentation will be removed and the skin will be cleansed with rubbing alcohol and water. The same cleansing process will be administered after the completion of testing.

Cuts – when shaving with the disposable razor, the possibility of a small cut exists. In the case of a cut occurring, the lab is equipped with a first aid kit, which includes plastic rubber gloves for the research (and participant if they want as well), hydrogen peroxide to sterilize the cut (cotton swabs to apply), and band aids.
to cover the cut.

**Emotional/Psychological:** All procedures will be given to the participant between the letter of information and consent, including the type of clothing that is necessary to complete this study. Participants, if they do not feel comfortable will not complete in the study. Furthermore, if the participant choses to participate, and begins to feel uncomfortable, they are free to withdraw from the study at any time, with no penalty. Constant communication of procedures will be relayed to the participant so they know exactly what is happening at all times, which will ensure participants feel comfortable in any physical or psychological/emotional state.

**Dual Roles:** This research in no way is related to your academic standing at the University. Your participation in this study is completely voluntary, and is not related to any course work, or University standing. Paul Leuty will deal with consent forms & recruiting, and you will have minimal, if any, interaction with Dr. Cort. Upon signing the consent form, you will be given a personalized subject code, which does not have any indication of your name, or your affiliation as a student of the university.

**POTENTIAL BENEFITS TO PARTICIPANTS AND/OR TO SOCIETY:**
Participants will be exposed to advanced exercise-biomechanics research practices which can benefit their awareness of personal biomechanics in activities of daily living. Furthermore, participants will experience the collection procedures of electromyography and motion capture as well as being briefly exposed to the Biodex for studying Isometric contractions which may be useful in future academics and/or careers. The scholarly community will be able to expand existing knowledge of fatigue on lower limb knee mechanics, as well as be introduced to a new manner of testing, which has never previously been studied. This research will open up many areas of research involving progressive fatigue, completing the same task from rest-fatigue, and examining the effects throughout the duration of the study.

**COMPENSATION FOR PARTICIPATION**
Each participant will receive a Kinesiology t-shirt, provided by the Faculty of Human Kinetics from the University of Windsor.

**CONFIDENTIALITY**
The testing sessions will take place within the Occupational Simulation and Ergonomics Laboratory at the University of Windsor. You will be assigned a randomly generated subject code known only to the investigators and therefore your identity can not be determined by anyone other than the investigators. Your personal information including name, age, and physical characteristics will be kept anonymous on all documents using the coding system. The information obtained in this study will be used for research purposes only and will be kept in a locked cabinet or stored on a password protected computer for a maximum of 5 years. There will also be no video recording or digital photos taken during the study.

**PARTICIPATION AND WITHDRAWAL**
You are being invited to volunteer in this study. If you choose to volunteer, you are free to withdraw from the study without any consequence at any time either before or during the testing sessions. If you choose to withdraw, all of your digital
data will be permanently deleted from the computers and all paperwork will be shredded. Withdrawal is no longer possible once the data is de-identified and amalgamated into a larger data set.

FEEDBACK OF THE RESULTS OF THIS STUDY TO THE PARTICIPANTS
Personal data sets and the final results of the study will be made available you if you are interested. You may obtain the results by providing the investigators with your email address at the time of testing, or by contacting one of the investigators at a later date. All collected email addresses will be kept confidential and only used for the purpose of sending out the final study results.

SUBSEQUENT USE OF DATA
These data may be used in subsequent studies, in publications and in presentations.

RIGHTS OF RESEARCH PARTICIPANTS
If you have questions regarding your rights as a research participant, contact: Research Ethics Coordinator, University of Windsor, Windsor, Ontario, N9B 3P4; Telephone: 519-253-3000, ext. 3948; e-mail: ethics@uwindsor.ca

SIGNATURE OF RESEARCH PARTICIPANT/LEGAL REPRESENTATIVE
I understand the information provided for the study Understanding the effects of progressive fatigue on impact landing force and knee joint mechanics, during the landing phase of continuous maximal vertical jump as described herein. My questions have been answered to my satisfaction, and I agree to participate in this study. I have been given a copy of this form.

______________________________________________
Name of Participant

______________________________________________
Date

______________________________________________
SIGNATURE OF INVESTIGATOR

These are the terms under which I will conduct research.
Appendix C

Lower Limb Health Questionnaire

Initial of first name: __________  Initial of last name: __________  Date:

________________________________

Age: ____________   Height: __________ ft. __________ in.    Weight (lbs):
_________________

Have you at any time throughout the previous 6 months had any discomfort, injury or disorders of the lower body, including your left & right hip, knee, ankle, foot (ache, pain, general discomfort, strain, sprain, break, numbness, injury, surgery) in:

<table>
<thead>
<tr>
<th>Hip:</th>
<th>Y</th>
<th>N</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee:</td>
<td>Y</td>
<td>N</td>
</tr>
<tr>
<td>Ankle:</td>
<td>Y</td>
<td>N</td>
</tr>
<tr>
<td>Foot:</td>
<td>Y</td>
<td>N</td>
</tr>
</tbody>
</table>

Have any troubles of the aforementioned areas affected your daily living in the past 6 months?

Y     N

Have any troubles of the aforementioned areas resulted in surgery?

Y     N
Appendix C

Returning Participant Questionnaire
In the past 48 hours have you experienced any muscle, tissue or joint pain, or discomfort, in any of the following area’s:

<table>
<thead>
<tr>
<th>Area</th>
<th>Y</th>
<th>N</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Foot</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Have you completed any strenuous physical activity (intense exercise exertions, such as athletic competition, or any personal training) in the past 48 hours?

<table>
<thead>
<tr>
<th>Y</th>
<th>N</th>
</tr>
</thead>
</table>


Get involved in Kinesiology Research!

We are seeking participants for a research study on **Progressive effects of fatigue while completing Maximal Vertical Jumping** and we want you to join!

**Are you:**

- Between the ages of 17 & 26?
- Free from, hip, knee, ankle and foot pain?

If so, you may be eligible to participate in a research study conducted by Dr. Joel Cort and his co-investigators, Paul Leuty and Chad Sutherland, (cortj@uwindsor.ca, leuty@uwindsor.ca, chads@uwindsor.ca). This study is completed across two sessions, and will require 4 hours of your time.

Please feel free to contact Dr. Cort or the co-investigators by email or telephone (519-253-3000 ext. 4277) if you would like to participate or want more information. This research has been cleared by the University of Windsor Research Ethics Board.
<table>
<thead>
<tr>
<th>NAME:</th>
<th>Paul Leuty</th>
</tr>
</thead>
<tbody>
<tr>
<td>PLACE OF BIRTH:</td>
<td>Stittsville, ON</td>
</tr>
<tr>
<td>YEAR OF BIRTH:</td>
<td>1990</td>
</tr>
<tr>
<td>EDUCATION:</td>
<td>South Carleton High School, Ottawa, ON, 2008</td>
</tr>
<tr>
<td></td>
<td>McMaster University, H.B.Sc, Hamilton, ON, 2012</td>
</tr>
<tr>
<td></td>
<td>University of Windsor, M.H.K., Windsor, ON, 2016</td>
</tr>
</tbody>
</table>